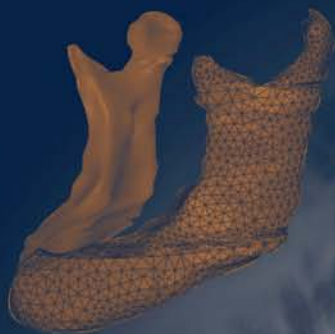




BIODENTAL ENGINEERING

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Biodental Engineering

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Preface

This book contains keynote lectures and full papers presented at *the I International Conference on Biodental Engineering*, held in Porto, Portugal, during the period 26–27 June 2009. The event had 5 invited lectures, 37 oral presentations distributed by seven sessions and 37 posters. The contributions came from 11 countries: Brazil, Colombia, France, Greece, Italy, Portugal, Romania, Spain, Turkey, United Kingdom, and United States of America.

The growing interest in the technology approach to medicine leads to the emergence of multidisciplinary research areas of great importance. From the wide range of issues that arise from scientific research urges the necessity to create spaces for thematic debate.

The dentistry is a branch of medicine with its own peculiarities and very diverse areas of action. The use of new techniques and technologies is currently the subject of great interest, and this conference was intended to be a privileged space for discussion among all stakeholders.

The purpose of the I International Conference on Biodental Engineering was to solidify knowledge in the field of bioengineering applied to dentistry.

The organizers would like to take this opportunity to thank, Faculdade de Engenharia da Universidade do Porto and Faculdade de Medicina Dentária da Universidade do Porto, all the sponsors, the members of the Scientific Committee, the Invited Lecturers and all Authors for submitting their contributions.

The Conference Organizers

Keynote papers

Collecting research data from clinical practice: How can informatics help?

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ABSTRACT: Evidence-based dentistry challenges the dental profession to base more and more of its clinical care activities on solid data regarding patient diagnoses, treatment and outcomes. However, systematic reviews in dentistry often suffer from a dearth of clinical studies. As a complement to large-scale, long-term clinical trials, many experts have advocated conducting clinical studies in dental practices. While this option presents some exciting possibilities and opportunities, it also faces significant challenges. Research data collected in clinical practice can be highly useful in dental research, because they are gathered in the settings in which care is typically rendered. As such, they can provide a highly credible source for information to guide clinical practitioners. On the other hand, conducting clinical studies in private practices faces obstacles related to practitioner training, data standardization, regulatory compliance and study design. Recent informatics initiatives can support dental field studies through standardization of data, a data collection and curation infrastructure, and quality assurance. In future years, informatics and health information technology is expected to play an increasing role in conducting field studies in dentistry.

1 BACKGROUND

Applying evidence-based dentistry principles during day-to-day patient care is an increasingly strong trend in dentistry. According to the American Dental Association (2008), evidence-based dentistry is “an approach to oral health care that requires the judicious integration of systematic assessments of clinically relevant scientific evidence, relating to the patient’s oral and medical condition and history, with the dentist’s clinical expertise and the patient’s treatment needs and preferences.” This “best evidence” is obtained from randomized controlled clinical trials, nonrandomized controlled clinical trials, cohort studies, case-control studies, crossover studies, cross-sectional studies, case studies or, in the absence of scientific evidence, the consensus opinion of experts in the appropriate fields of research or clinical practice (Ismail & Bader, 2004). Such evidence is typically synthesized in systematic reviews that are based on a comprehensive analysis of the literature.

However, as many systematic reviews in dentistry point out (Ahovuo-Saloranta et al., 2008, Esposito et al., 2006, Bolla et al., 2007), the number of published studies is often insufficient to draw conclusions with the required confidence to recommend changes in or confirm the appropriateness of practitioner behavior. In light of this situation, the National Institute of Dental

and Craniofacial Research (NIDCR) recently committed \$75 million to establish three practice-based research networks (PBRNs) in general dentistry that will answer questions raised by dental practitioners in the everyday practice of dentistry (Pihlstrom & Tabak, 2005). The objective of the PBRN initiative is to accelerate the development and conduct of clinical trials and other clinical studies of important issues concerning oral health care related to general dental practice. The focus of research within the PBRNs is on relatively short-term clinical studies, with emphasis on comparing the effectiveness of various oral health treatments, preventive regimens and dental materials.

While conducting research studies in dental offices is not novel, it definitely has not been common to date. The establishment of the PBRNs therefore presents not only significant opportunities, but also sizable challenges. Dental offices in the United States typically deliver care through a highly efficient and systematic process. While conducting research in private practices is an important goal, the question must be raised how research can be most efficiently and effectively be integrated into the practice workflow. Specifically, this paper focuses on what role information and information technology can play in meeting this challenge. We briefly review the functioning and operations of dental PBRNs before we turn to the specific informatics issues they face.

2 PBRNs IN THE UNITED STATES

In April 2005, the NIDCR funded three practice-based research networks to encourage the conduct of clinical research in dental practices. The three networks are distributed across geographic regions in the United States: the Northwest Practice-based REsearch Collaborative in Evidence-based DENTistry (PRECEDENT) (2009), administered through schools of dentistry at the University of Washington (UW), Seattle, and Oregon Health and Science University (OHSU), Portland; the Practitioners Engaged in Applied Research and Learning (PEARL) Network, administered by the New York University College of Dentistry (2009); and the Dental Practice-Based Research Network (DPBRN) (2009), administered by the University of Alabama at Birmingham.

To date, the networks have enrolled a total of over 1,400 practitioners, with 1,100 participating in the DPBRN alone. As of April 2009, over 45 studies were either completed or in progress. Studies cover a diverse range of subjects, such as salivary markers and genetic factors in caries risk assessment, pulp capping with mineral trioxide aggregate vs. calcium hydroxide, treatment outcomes for noncarious cervical lesions, periodontal diagnosis and referral, blood sugar testing in dental practice, and Internet-based tobacco cessation interventions. Three studies are currently being conducted across all three networks, including a case-control study of osteonecrosis of the jaws.

Dental offices that become members of one of the networks in order to conduct in studies typically go through the following process. First, the office staff is oriented about the responsibilities and activities necessary to conduct research studies. Practitioner-Investigators (P-Is), i.e. the individuals in the practice who will conduct one or more studies, are trained in the principles of good clinical research, ethical issues and the provisions of the Health Insurance Portability and Accountability Act (HIPAA) (DeRouen et al., 2008). (HIPAA is a federal law designed to protect the privacy and confidentiality of patient information.) An introduction to research methods and the conduct of research studies completes the P-I training.

Practices conduct the studies with the assistance of Coordinating Centers (CCs). CCs are clinical research organizations with in-depth experience in conducting clinical studies. They provide the necessary forms, for instance for patient consent and data collection; orient the practice staff about study details, such as eligibility criteria and data collection processes; monitor the completion of case report forms; and provide quality assurance.

While some studies and practices use only paper-based case report forms, electronic data

gathering through Web-based systems is becoming increasingly common (Pace & Staton, 2005). In one scenario, study personnel enter data for each study participant through a Webpage, and the data can immediately be checked for completeness and accuracy. However, data capture solutions that employ notebook computers, tablet PCs, personal digital assistants are also available (Pace & Staton, 2005). Electronic systems allow the use of transparent decision algorithms and improved data entry and data integrity. These systems may improve data transfer to the central office as well as tracking systems for monitoring study progress.

Participating in a PBRN means that practices must overlay the work process related to research on top of their practice operations. Clearly, doing so requires significant effort to make sure the two processes function harmoniously and with a minimum of extra effort for practice team members. However, in practice the existence of two separate work processes and data gathering methods tend to create duplication and extra work. For instance, at present none of the PBRNs is reusing data from dental practice management systems (PMS) in participating practices. Before we turn our attention to the types of strategies that can be used to mitigate these problems, it is worthwhile to take a look at the state of computerization in dental practice.

3 INFORMATICS ISSUES IN PBRNs

As the experience in medicine has shown, the information technology (IT) infrastructure has a key role in determining the success of clinical research projects (Koop & Mosges, 2002, Sung et al., 2003). It is therefore useful to consider how the nascent dental PBRNs can take advantage of the existing IT infrastructure in dental practice in order to be as effective and efficient as possible.

In the United States, computerization in dental practice is relatively widespread. As of 2006, 92.6% of all dentists used a computer in their office (American Dental Association Survey Center, 2007). However, the degree to which computers are used in support of administrative and clinical processes varies widely. Patient accounting and billing, processing insurance forms, and scheduling are the most frequently reported uses of the computer. Internet access is installed in 72.8% of all practices. The clinical use of computers, on the other hand, is less common. A comprehensive study conducted by our Center (Schleyer et al., 2006) showed that 24.6% of all general dentists have a computer in the operatory and 1.8% maintain completely paperless patient records.

Particularly relevant for PBRNs is the degree to which dentists use computers in the clinical

context. The larger the degree to which clinical information about patients is managed on the computer, the greater are the opportunities to capture that data for research purposes. Data from our study (Schleyer et al., 2006) show that the amount of clinical information stored on computers varies widely (see Fig. 1). Clinical information with a strong connection to practice management (i.e. appointments, treatment plans and completed treatment) is stored most often on the computer; images and clinical charting (radiographs, photographs and intraoral charting) less often; and other clinical information (e.g. diagnoses, dental and medical history, and progress notes) least often.

Implementing clinical research in a busy dental practice faces a number of significant hurdles. Most dental clinicians are unfamiliar with the process of conducting research. Collecting and managing research data must be implemented as a process alongside the ongoing management of clinical data. Associated tasks, such as determining eligibility, consenting patients, recording adverse events, scheduling follow-up visits and performing audits, must be integrated into the practice workflow.

How successful dental practices will be in conducting clinical research depends partly on how well the research process is integrated with practice operations overall. The less overhead a PBRN creates, the more likely it is that its participating practices will perform high-quality research. Given the

relatively widespread use of computers in dental practice, we should consider how the existing IT infrastructure can be leveraged to facilitate the research process. In the next section, we examine this issue by looking at two facets of integration: the integration of research data management, and of research workflow with dental practice management systems.

4 A PROPOSED STRATEGY FOR LEVERAGING INFORMATICS

We recommend integrating the research process with practice operations in two successive stages: *Data Integration* and *Workflow Integration*. The main objective of *Data Integration* is to integrate research data entry and management progressively with the PMS already in use in the practice. At the highest level of *Data Integration*, all data entry and management functions are performed through the PMS. However, the practice must still perform other research-related functions manually, such as screening patients for eligibility, assuring that patients are scheduled for the correct number and type of visits, and determining whether to activate a specific research protocol in the practice. The next level of integration, *Workflow Integration*, encompasses *Data Integration*, and embeds most of the knowledge that is required to initiate and conduct research in a dental practice in the IT infrastructure. At its highest level, *Workflow*

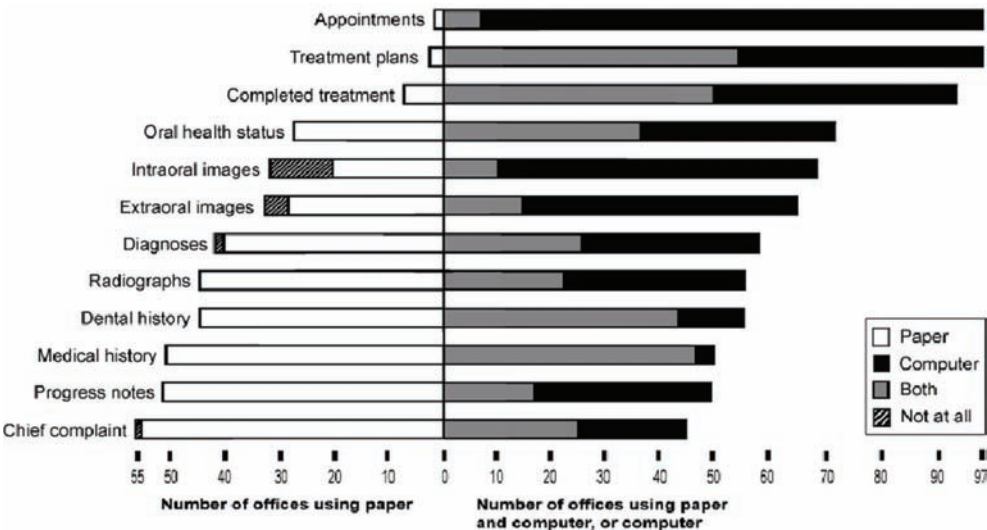


Figure 1. Storage of major clinical information categories on paper/computer, sorted by utilization of computer-based storage in descending order (Paper = information stored only on paper; Computer = information stored only on the computer; Both = information on paper duplicated on the computer; Not at all = information not recorded at all).

Integration assists the practice in deciding whether to become part of a PBRN study, in identifying the most promising research protocols to initiate, and in managing and conducting each study itself.

In this manner, the IT infrastructure assumes much of the operational workload in conducting research, and frees up the practitioner-investigator and other personnel to concentrate on value-added research tasks and patient care. Successful IT integration also relieves some of the financial burden that a dental practice may have to absorb by participating in a PBRN study.

Below, we illustrate the somewhat abstract concepts of *Data Integration* and *Workflow Integration* using brief descriptions and, where appropriate, scenarios. Both types of integration are separated into three successive levels.

4.1 *Data Integration*

We first describe the three levels of Data Integration. *Level 0 (Practice and Research IT Infrastructure Separate)* is the logical starting point in which research data and practice data are managed separately. Typically, the dental office will continue to maintain practice data on its PMS, while research data may be collected through paper forms, tablet computers and other mechanisms. Because this method can not take advantage of electronic, study-related data that already exists in the practice, it creates significant overhead and should be avoided.

Level 1 (Research Study Takes Advantage of Existing Electronic Patient Data) uses electronic patient data that are already stored in the practice PMS. A sample scenario for this level of integration begins with a patient who is not yet enrolled in a clinical study. After Mr. Brown, a patient of record, has arrived in the practice, the receptionist, Sally, tells him about a research study with a new antibacterial mouth rinse that the office just started. Mr. Brown is interested in participating in this study. After determining that he is eligible, Sally starts the research data management software (RDMS) on a tablet PC and begins enrollment of the new participant. She activates the “Search” feature in the software, enters Mr. Brown’s last name and initiates the search. The tablet PC then queries the PMS database and returns a list of matches. Sally selects the correct patient record and hands the tablet PC to Mr. Brown. He completes the enrollment process, including informed consent. These forms have been pre-populated with Mr. Brown’s demographic information, saving time and reducing errors. During enrollment, the research data management application retrieves study-related data from the PMS, such as part of the health history and the latest PSR scores. Once

Mr. Brown is seated in the dental operatory, the dentist, Dr. Smith, performs the treatment planned for Mr. Brown that day. When Dr. Smith is ready to begin gathering study-related data, he invokes the RDMS through a menu item in the PMS. (The dental IT industry has been using this technique, called “bridging,” for many years to connect separate software applications.) During bridging, Mr. Brown’s record is automatically selected in the RDMS. Dr. Smith then records study-related data and saves them. The complete set of study-related patient data, both those read from the PMS and those newly acquired, are then transmitted to the central research database.

Level 2 (Research Study Data Collection and Management Fully Integrated into the PMS) takes the integration of research data management with the PMS from the database level to the user interface level. At this level, the PMS and the RDMS appear as one homogeneous application to the user; the RDMS ceases to exist as a separate application. The sample scenario again begins with Mr. Brown being enrolled in a research study. However, Sally selects Mr. Brown’s record in the PMS, and then simply clicks on the “Research Data Tab.” On this tab, Sally selects the study in which to enroll Mr. Brown. Mr. Brown then completes the enrollment process on the tablet PC that Sally hands him. In the operatory, Dr. Smith enters patient care-related data into the PMS as usual. To begin research data collection, the practitioner simply switches to the “Research Data Tab” and records study-related data for this visit.

Data Integration facilitates research data entry and management; however, it does not address the integration of research as a process into the practice workflow. That is the goal of *Workflow Integration*, which, like *Data Integration*, has three levels.

4.2 *Workflow Integration*

At the lowest level of *Workflow Integration*, practice workflow and research workflow are, again, considered separate processes. Beyond this initial stage, *Level 1 (Research Protocol Management Integrated into the PMS)* integrates knowledge about study protocols into the practice’s PMS. This knowledge consists of (1) the data specific to each research protocol; (2) the number and types of contacts each study participant has with the P-I during the research protocol (such as surveys and clinical examinations); and (3) other rules contained in the research protocol (such as number of arms of the study and drug dosages). A software module that “plugs into” the PMS, the *Research Protocol Agent*, encapsulates this knowledge. In addition to providing data entry functions,

the PMS generates alerts, reminders and status messages for study participants.

In the sample scenario, Mr. Brown is enrolled in the study as described above. As he is seated in the dental operatory, the dental assistant, Mary, locates his record in the PMS. Since the initial clinical data collection for the study is also due at the first visit, the *Research Protocol Agent* alerts Mary that data for Form 1 of the research study should be collected. After finishing patient treatment, Dr. Smith, reminded by Mary, completes the research data collection. At the conclusion of the appointment, Mary schedules Mr. Brown for another appointment in two weeks. At the same time, the *Research Protocol Agent* alerts Mary that the first follow-up appointment for the study must occur six weeks from the current date. The software suggests a date and time, Mary accepts the suggestion, and schedules the second study appointment. The practice typically sends reminder postcards a few days before patients' appointments. In the case of the upcoming study appointment, a letter must be generated that provides Mr. Brown with specific instructions. The *Research Protocol Agent* generates this letter through a batch process a few days before the appointment, and the receptionist mails the letter that was printed. Based on clinical data gathered at the following study visit, Mr. Brown is enrolled in the intervention arm of the protocol. The *Research Protocol Agent* automatically selects the appropriate forms for this arm of the study, and continues the support for Mr. Brown's flow through the research protocol until he is discharged from the study.

Level 2 (Research Process Integrated into Practice Workflow) integrates the research process maximally with the practice workflow and operating processes. At this level, routine and administrative tasks are maximally offloaded to the information technology infrastructure, so that the P-I and other practice personnel can concentrate on core research activities. At this level, a *Research Support Agent* provides support for most research functions, beginning with the enrollment of a practice into the PBRN to its termination. The purpose of Level 2 of *Workflow Integration* is to allow practices to join the research network, regardless of their geographical location, with minimal effort and travel, and facilitate their participation in research trials using a set of computer-based tools that integrate with the PMS already operating in the practice.

In the sample scenario, Dr. Smith has been interested in conducting clinical research in his practice for some time. After seeing a solicitation for practices interested in clinical research in a newsletter from the American Academy of General Dentistry, he decides to investigate further. He views an

informational module on practice-based research provided on the PBRN Website. His practice profile indicates that he would be a suitable participant in the research network. He initiates his enrollment directly through the informational module. Based on the information he submitted, his application is approved, and a research coordinator contacts Dr. Smith a few days later to welcome him to the network and discuss the next steps. First, Dr. Smith downloads the *Research Support Agent* software, which includes a plug-in for Dentrux, the practice management system in use in his practice. Four study protocols are currently active and available for Dr. Smith's type of practice. After viewing summaries of the protocols, Dr. Smith would like to find out which protocol(s) would be most suited to his patient database. Since Dr. Smith's practice is completely paperless, the *Research Support Agent* provides a function that automatically scans each patient record within Dentrux for inclusion and exclusion criteria for the protocols under consideration. Dr. Smith initiates the scanning function. After a few minutes, the *Research Support Agent* reports that Dr. Smith's patient database is well suited for two protocols.

After discussion with his staff, Dr. Smith decides to participate in a trial for a dental implant with a novel coating that supposedly results in improved osseointegration. The *Research Support Agent* downloads the protocol in a structured, machine-readable format. After downloading the protocol, the *Research Support Agent* offers Dr. Smith the option of contacting patients of record eligible for the study. Dr. Smith accepts the suggestion, and the *Research Support Agent* e-mails patients with e-mail addresses in the PMS information about the study. For patients without e-mail addresses, the software generates hardcopy letters which are sent by mail.

A new patient, Charles Byers, registers in the practice the next day. After the treatment plan for Mr. Byers is completed, the research plug-in alerts Dr. Smith about the fact that Charles would meet the inclusion criteria for the newly initiated study. Dr. Smith briefly reviews the study with Mr. Byers. A few days later, Mr. Byers calls back to tell Dr. Smith that he would like to participate in the study. On several subsequent visits, study-related data is collected and transmitted to the central research database. The data transmission includes a variety of textual and numerical clinical data, as well as several radiographs of the implant. After the third appointment, a slight change in the study protocol is made by the protocol office. Dr. Smith receives an e-mail message about the change, and the *Research Support Agent* automatically downloads an updated version of the protocol. Charles concludes his participation in the study as scheduled two years after his initial appointment.

The scenarios illustrate several benefits of progressively integrating the research process with practice operations. At Level 1 of *Data Integration*, duplicate data entry is minimized or eliminated. However, the user must still work with a separate software application for research. At Level 2, research data management is integrated into the PMS, reducing time and cognitive load needed to enter research data. At Level 1 of *Workflow Integration*, research personnel are freed up from having to implement a system to track and manage activities associated with individual studies, such as scheduling and selecting correct data entry forms. Finally, at Level 2 of Workflow Integration, the IT infrastructure encapsulates most of the knowledge needed to initiate and conduct research in dental practice. At this point, the research staff can focus almost exclusively on value-added research tasks, such as decision-making, data collection and analysis.

5 CONCLUSION

Implementing practice-based research networks in dentistry is a challenging but potentially highly rewarding endeavor. As the experience in some medical PBRNs has shown, information technology can be leveraged quite effectively in support of PBRN operations (Kho et al., 2007, Stephens & Reamy, 2008). For instance, within ResNet, a PBRN of 17 primary care sites in Indiana, the Regenstrief Institute screened more than 18,000 patients and enrolled more than 6000 study subjects in 5 years, missing less than 2% of potentially eligible patients by using data from an electronic medical record system (Kho et al., 2007).

A key premise of the discussed approach of integrating research into the practice's IT infrastructure is to reduce the procedural and cognitive overhead, so practice-based researchers can concentrate more fully on conducting research itself, rather than managing it. Successful integration minimizes both the financial impact and the impact of research tasks on the normal patient care and office workflow. Our suggestions are most appropriate for practices with a high degree of IT literacy, and with a well-developed and well-used IT infrastructure. The more clinical patient data are available electronically in those practices, the larger is the degree to which these data can be leveraged for research. Such highly computerized practices currently make up only a very small proportion of all practices, both in dentistry (Schleyer et al., 2006) as well as in medicine (Burt & Sisk, 2005, Jha et al., 2006, Andrews et al., 2004). However, if historical trends are any indication, the sophistication of the IT infrastructure in dental practice will

continue to increase. In this way, a "central nervous system" for clinical research in practice can evolve. This will eventually develop into a tremendous asset for improving the oral health of patients in the U.S. and elsewhere.

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Human temporomandibular joint simulation

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ABSTRACT: Finite element simulations of the human temporomandibular joint (TMJ) open a new insight in the understanding of how this joint works and why there is a high prevalence of TM disorders (TMD). Several works have analyzed the joint under different pathologic situations, also studying its response under some surgical procedures.

1 INTRODUCTION

The temporomandibular joint (TMJ) is one of the least studied joints in the body, in spite of the large patient population and the significant morbidity related to a large number of TMJ disorders. Epidemiological studies have reported that 20–25% of the population exhibit symptoms of TMJ disorders, while patient studies show that 3–4% of the population seek treatment. The most common TMJ disorders are pain dysfunction syndrome, internal derangement, arthritis and traumas. Collectively, current treatments are not able to fully address severe TMJ disorders (Detamore et al., 2007).

Bioengineers worldwide have made major strides in modeling the TMJ, with major contributions from groups in Japan, the Netherlands, Spain and Switzerland (Detamore et al., 2007). Many finite element (FE) analysis have been reported to probe the biomechanical status of the TMJ. The first finite element models were bidimensional (Chen & Xu, 1994, deVocht et al., 1996). In these models the biomechanics of an oversimplified human TMJ was analyzed during the normal movement of the jaw, and although no reliable material properties were available for the different tissues, it was predicted that there was a nonlinear relationship between the maximal stresses that appeared in the tissues and the stiffness of the disc. The first three dimensional finite element model was constructed by Nagahara et al. 1999. Koolstra & van Eijden (1999) developed a 3D model that include the masticatory muscles. Later, Beek et al. (2000) reported that the stresses were located in the intermediate zone of the disc during clenching. The results of these studies suggested that the disc plays an important role in distributing and absorbing loads acting on the joint. The applied models, however, were (quasi-)static and, therefore, the applicability of their results is limited to situations where the jaw hardly moves

(e.g., clenching) (Beek et al., 2001). Posteriorly, Hu et al. (2003) introduced in a simplified model of the joint, the fibrocartilaginous layers that cover the articulating surfaces of the joint, concluding that these prevent the disc and bone components from high stresses. Donzelli et al. (2004) presented a 3D model of the joint, where the disc was considered as an isotropic poroelastic material.

The first model that took into account the presence of the collagen fibers inside the tissue, and the transversely isotropic character of the solid matrix was proposed by Pérez del Palomar & Doblaré (2006), concluding that the introduction of collagen fibres in the biphasic behavior of the articular disc implies for a prescribed displacement not only an increase of the pressurization in the tissue, but also higher stresses in the anterior and posterior bands, as well as in the lateral zone of the disc. Following this idea, these authors also simulate the dynamically opening movement, lateral excursion of the jaw (Perez-Palomar & Doblaré, 2006) proposing different mechanisms of damage of the joint, and also the most common pathology such as the anterior disc displacement.

Therefore, the development of computational models that can mimic the behaviour of the TMJ is a challenging goal for biomechanical engineers. In this paper a summary of the model proposed by Perez del Palomar & Doblaré is presented.

2 MATERIAL AND METHODS

2.1 Finite element model

The geometric data for the model was obtained from a healthy male patient. The contours of the cranium (temporal bone) and the mandible were obtained from CT images and soft tissues contours were constructed from MR images. Bones were considered to be rigid since they are much stiffer than

the relevant soft tissues and were automatically meshed using I-deas v.9. The deformable parts of the joint, the articular disc and the ligaments were manually created and meshed. The definition of the contours of the disc was based on MR images with splines detected semiautomatically by means of a custom-design code, taking into account that the disc adapts its shape to the surface of the temporal and the condyle. The temporomandibular ligament could be located from the MR images, and then manually generated. But the attachment position of the joint capsule was determined based on the observed disc position and shape of bony components and anatomical knowledge of the TMJ. The finite element meshes of discs and ligaments were constructed in I-deas v.9 using eight-node brick elements.

Finally, the retrodiscal tissue was not meshed. This tissue is composed partially of ligaments connecting the disc to the temporal bone and the condyle, but the primary composition is made up of a fairly loose, spongy material. Thus, this tissue was replaced by a set of springs with a similar orientation. These were defined between the posterior part of the disc and the temporal bone, and their stiffness was computed from the stiffness of the retrodiscal tissue. The finite element model constructed included therefore the mandible, the temporal bone, the two temporomandibular joints, the articular discs, the lateral and medial ligaments, the external lateral ligaments and the retrodiscal bands (Figure 1).

2.2 Material models

The soft elements were treated as hyperelastic materials. A fiber reinforced porohyperelastic model was introduced for the discs, where collagen fibers run along the mediolateral direction in the bands and the anteroposterior direction in the intermediate zone (Shengyi & Yinghua, 1991, Berkovitz 2000,

Perez-Palomar & Doblare, 2006). The constitutive model for the disc was implemented as a user subroutine in the general-purpose finite element code ABAQUS v.6.4; this model is extensively detailed in (Perez-Palomar & Doblare, 2006). For the ligaments, a Neo-Hookean hyperelastic model with $C1 = 6$ MPa (Weiss et al., 1996) was used. Finally, although the stiffness of the bilaminar zone is not well known, some researchers (Tanaka et al., 2002) reported that it is around 1.5 MPa. For our calculations, we considered that the effective area of the bilaminar zone was 50 mm^2 and its length 15 mm, therefore, the equivalent stiffness of the equivalent set of springs was 6.5 N/mm .

With regard to the interaction between the elements, nine contact pairs were needed to define their interaction. A friction coefficient of 0.0001 was considered for all the interactions because of the existence of synovial fluid (Tanaka et al., 1994, 2001).

2.3 Loading and boundary condition

To simulate both active and passive responses of the muscles, connector elements were used in ABAQUS (HKS inc. Pawtucket, RI, USA, 2003). These elements allowed including a nonlinear stiffness depending on their stretching and also a damping coefficient to take into account both their passive resistance and their viscous behaviour. In addition, these elements can exert a follower force that can be activated at a specific time reproducing the active response of the muscle. These connector elements were defined between their insertion points (Gal et al., 2004, Langenbach et al., 1999, Peck et al., 2000).

Several cases were considered. The movement of the mandible during opening, closing, chewing can be analyzed. Also, the changes on the stress distribution when the disc is for instance anteriorly displaced can be studied. Moreover, it is known that the contact between the teeth affects the response of the joint, therefore the influence of different premature contacts was analyzed.

3 RESULTS

The movement of the mandible can be depicted by contracting the openers to different levels. In Figure 2 this movement is shown for a healthy TMJ and one affected of an anterior disc displacement. It can be seen how the maximum displacement of the mandible is smaller when there is a pathology of the joint, and how this kind of disease can introduce a nonsymmetric movement of the jaw.

As mentioned in the previous section, it is known that an improper contact between the teeth can

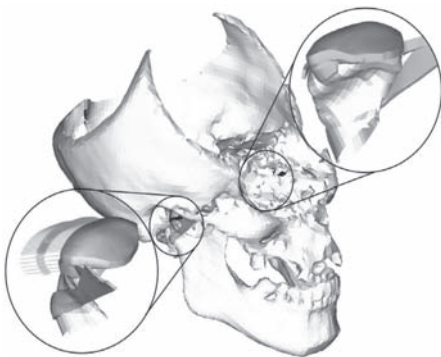


Figure 1. Finite element model of the TMJ.

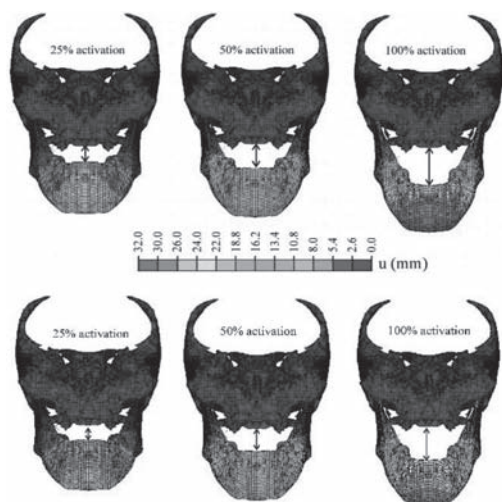


Figure 2. Movement of the mandible for different activation level of the openers. In the first row the results for a healthy patient is depicted, while in the second the movement for a unilateral anterior disc displacement is shown.

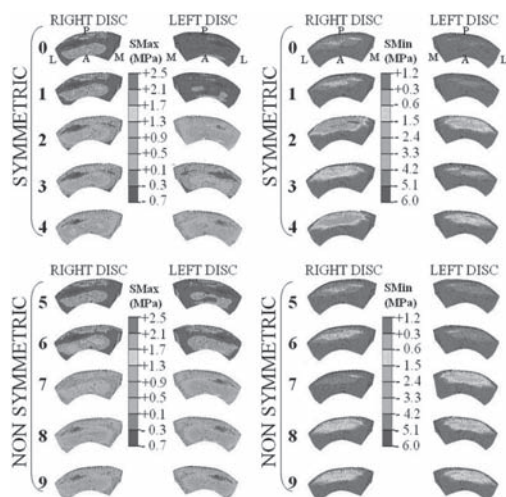


Figure 3. Distribution of maximum and minimum stresses in both discs (coronal view) for the different analyzed cases. Cases 0–4 represent symmetric patterns while cases 5–9 nonsymmetric ones where different teeth contacts are simulated. For more details see Perez del Palomar et al., *Annals of Biomedical Engineering*, 6(6), 1004–1023, 2008.

affect the tensional behaviour of the components of the joint. In Figure 3, the maximum and minimum principal stresses in both discs are presented for different teeth-contacts.

4 DISCUSSION AND CONCLUSIONS

Potential applications of accurate and validated FEMS of the human TMJ are vast. Such models can allow characterization of the thresholds of normal function and parafunction, which may help to explain the etiology of TMJ disorders. The direct applications of these computational techniques are to track the 3-D motions of the joint and to note the distribution of the stress-fields inside the TMJ in prescribed situations. These models may be used to evaluate force-fields that accompany different activities, e.g., talking, chewing, bruxism or whiplash, an to predict which activities might lead to tissue damage. This would also be valuable to clinicians by assessing loads in the TMJ during various procedures. For surgeons, one implication could be to serve as a guide to the biomechanical environment and functionality of the joint before and after surgery, for example to evaluate implants performance. Another application of FEMs of the human TMJ could be in the understanding the cause of functional adaptation and growth.

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Regular papers

Biomechanical study of changes in buccal bone structure induced by expansion screw

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ABSTRACT: Malocclusion problems frequently lead to dysfunctions in the jaw joint and alterations in the chewing, speech and breathing. Patients should be treated precociously for proper aesthetic improvements through bone remodeling and a better architecture of the dental arches. Techniques for correction require the installation of a mobile device in the mouth that, through the drive of a screw, imposes gradual displacements and forces a new balance position. It is aimed to reach a less traumatic application of forces, attenuating painful symptoms such as chronic headaches, bone partial necrosis and even loss of dental elements. This work seeks the quantification of the compression forces generated by these devices and a follow up of bone growth and remodeling. With this purpose, from the data obtained from in vivo experiments, the development of a biomechanical model of the dental movement including bone remodeling stimulation using finite element method is searched.

1 INTRODUCTION

Orthodontics searches for reaching occlusal equilibrium and the functionality of the stomatognathic system: bones, teeth, periodontal ligament, tongue, temporomandibular joints (TMJ) and other oral biological structures, that can happen during the growth of a person. When this equilibrium is not reached, morphological changes, known as malocclusions, occur. Malocclusion problems resulting from the maxilla's narrowing are commonly found in the population. They lead to dysfunctions in the jaw joint and alterations in chewing, speech and breathing. Patients should be treated precociously so that the results allow for proper aesthetic improvements. Steady dental position and a remodeled bone contour are desired through a functional occlusal gear for attaining normality. These bone growth disturbances cause an unfavorable esthetic effect that impacts in the self-esteem of the patients. Malocclusion patients, mainly children and teenagers, as well as young adults, need orthodontic and orthopedic treatment to open the maxillary suture and bone remodeling of the palatal contour and a better architecture of the dental arches. The orthodontist should know the level of applied forces in the oral correction devices and the clinical follow up of the treatment. The treatment involves reactions of organic structures, starting with inflammatory process, prevention pain and irreversible injuries to live tissues as well

as uncontrolled bone resorption and lost of tooth elements (Proffit, 1995; Burst & MacNamara, 1995). The connection between physical phenomena and organic reactions, in addition to knowledge of the material properties, is essential to get the expected results from the treatment. One of the techniques for correction of this pathology requires the installation of a mobile device in the mouth that allows bone remodeling, with a bone resorption area and deposition of neoformed bone tissue. It is necessary to use a methodology that correlates the imposed displacement with the bone structure deformation, cortical as well as trabecular, and the impact absorption by the mucosa, quantifying the level of forces and stresses reached. The purpose is to arrive at a less traumatic application of forces, attenuating painful symptoms such as chronic headaches, partial necrosis of the bone and even the loss of dental elements. The mechanisms of the dental movement phases and bone remodeling should preserve the biological and functional integrity of the stomatognathic system, not only the dental elements but also the occlusal gear, the periodont and the temporomandibular joint, as well the aesthetic and individual characteristics of the patients (Sandy et al., 1993). Expansion screw is the main orthopedic accessory used to allow bone remodeling in the dental arch with objective to correct deviations in the shape and boundary of the buccal bone and to reach an adequate dental occlusion (Haas, 1965; Silva Filho, 2007).

2 OBJECTIVE

The objective of this work is to quantify the distributed stresses in the live structures involved in the dental movement and bone remodelling during orthodontic/orthopedic treatments using expansion screw apparatus as well as the force levels reached in the mucosa and bone. The data obtained from *in vivo* experimental measurements are compared with the results obtained in simulations using Finite Element Method.

3 MATERIALS AND METHODS

Three experiments were developed. The first one was to measure the level of stresses on the oral structures from the use of expansion screw apparatus in the jaw. In the second experiment, the measurements were obtained from an expansion screw apparatus acting in the maxilla. In the third test, a bench was assembled to determine the Young's Modulus of a palatal mucosa, in this case a tissue of pig oral mucosa. The results of the measurements, validated through the comparison with technical literature, became input for the mathematical biomechanical model of the dental movement due to the bone growth stimulation using expansion screws. The collected data was tabulated and force versus displacements and force versus time graphs plotted. Figure (1) shows the *in vivo* measurement device ("Flex Force System").

Peak pressure values were obtained. The output was analyzed, and after noise filtering, registered

as the measured force level. The measurement were obtained directly from the patient after driving $\frac{1}{4}$ of turn during movements of chewing, swallowing and pressure of the tongue over the incisor reposition springs, in the anterior region and also in the molar region. The objective of the second experiment was to determine the forces acting in the intermaxillary suture, right and left sides, through the measurement of the pressures transmitted to live tissues in the points of compression and stress due to the apparatus accessories. The same method of the previous experiment was followed. The sensors were adapted in predetermined points and fixed by utility wax. Figure (2) shows the adaptation of the sensor in the apparatus and part of the electronic circuit of the second experiment, illustrated in the Figure (1).

The objective of the third experiment was to compare the measurements of Young's Modulus of a pig palatal mucosa with data from the literature of toothless fibromucosa that were using prosthesis. Mucous removed from pig palate with different ages were tested in the stress bench shown in the Figure (3).

To compare the results, it was also performed an *in vitro* experiment, simulating the load conditions



Figure 1. Force measurement system and data acquisition bench.

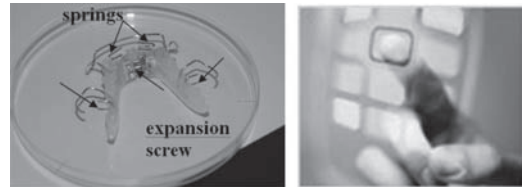


Figure 2. Sensor adapted and part of the electronic circuit assembled for the 2nd experiment.

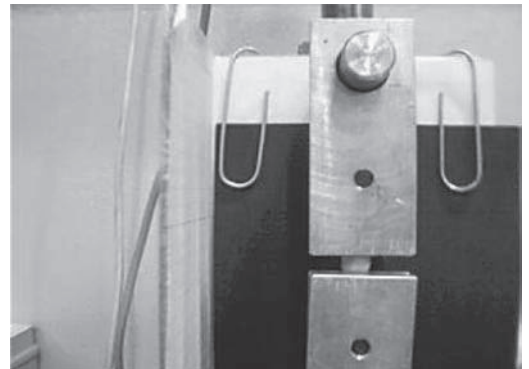


Figure 3. Stressing of a pig palatal mucosa (frontal view) developed by Martins et al., 2006 (FEUP).

by the action of the expensor apparatus causing stresses on the jaw live tissues.

The mechanical properties of the biological tissues are only approaches and the mathematical model to be applied tends in general to describe its behaviour in a limited way, giving answers related to a specific load condition (Özkaya & Nordin, 1999).

The numerical model was constructed using the geometric model of a human jaw, obtained from a computerized tomography with digital images, in partnership with Faculdade de Engenharia do Porto (FEUP). The computerized tomography of the face is showed in the Figure (4).

An example of the geometric model prepared using the commercial program Solid Works is showed in the Figure (5), with a model of the apparatus with expansion screw in the jaw.



Figure 4. Computerized tomography of the face.

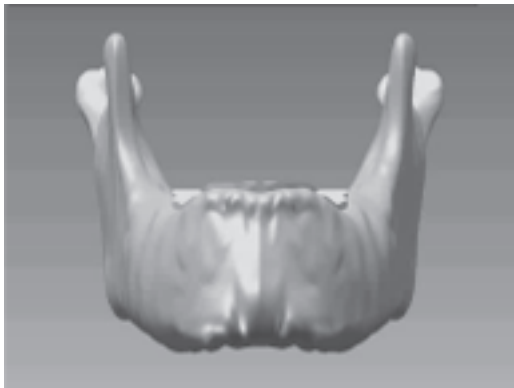


Figure 5. Geometric model of the jaw (Solid Works).

The geometry adopted to solve the problem includes the acrylic of the expensor device, the mucosa and the cortical and medular bones.

In the simulated model, the apparatus, bone and other biological tissues were considered as usual linear-elastic, homogeneous, isotropic materials, although these biomechanical materials, in fact, have self-repair capacity and are able to adapt to the mechanical request changes, modifying their mechanical properties (Özkaya & Nordin, 1999).

The following mechanical properties were attributed to the material of the geometric model: Young's Modulus and Poisson coefficient, respectively. For the acrylic apparatus they were 2400 Mpa and 0.35 respectively, for the mucosa 0.98 Mpa and 0.4 and for the bone 13700 Mpa and 0.3.

Taking the geometric model, it was developed the discretization of the problem and generated the mathematical model. To reduce the size of the model and in consequence less hardware demand, it was considered cranium symmetry of the right and left sides.

The mesh was created using the pre-processor from *Abaqus 6.7* software. 3D elements, tetraedric type, were utilized. This element, among the availables for usage, is the one that better adapts to the irregular boundary of the problem. The generated mesh can be seen in the Figure (6).

Afterwards, the numeric model was created, with respective boundary conditions. It was imposed displacement restrictions, which were vertically and laterally placed in the descent of the coronoid process and anterior-posterior of the condilo.

Because of the symmetry of the problem it was necessary to limit the lateral displacement in the posterior region of the jaw. Figure (7) details the boundary condition of the model: displacement restrictions of the bone and mucosa, and the displacement limits of the apparatus.

For the apparatus, it was imposed a displacement of 0.1 mm, that corresponds to $\frac{1}{4}$ of turn to open the expansion screw.

To typify the contact between the involved parts, it was created a structural rigid tie between



Figure 6. Generated mesh using the software *Abaqus*. From left to right, the expansion device, mucosa and jaw bone.

the mucosa and the bone, to avoid relative displacement between the structures, and a contact tie between the mucosa and the apparatus (without friction).

4 RESULTS AND DISCUSSIONS

The pressures in the structures simulated in the model (expansion screw apparatus—mucous—bone) reached good approach with the ones measured in the in vivo experiments. The mesh and pressure contours are shown in Figure (8).

In accordance with clinical observation and data from the literature, the region with bigger levels of stress coincides to the theory of dental movement from orthopedic expansion. Table (1) lists the results of the three experiments in comparison with the literature data and respective authors.

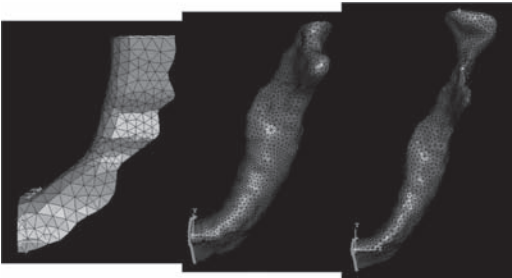


Figure 7. System boundary conditions. In the left, the apparatus limit displacement in x axis. In the middle, the mucosa movement restriction in the axis x referred to the imposed symmetry. In the right, in the anterior part, the bone movement restrictions in the axis y and z. In the posterior part, the restriction movement in the axis x was considered.

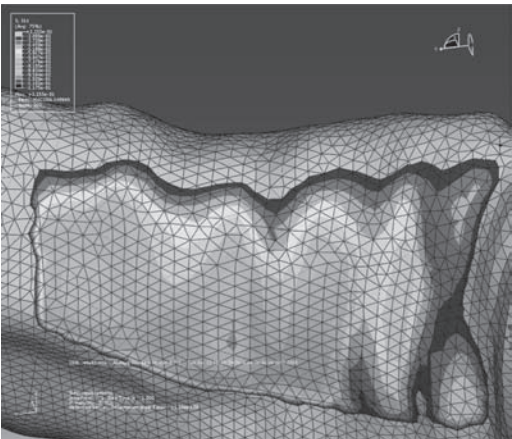


Figure 8. Mesh with isocurves of pressures.

Table 1. Applied forces from experiments in comparison with literature data.

Region	Data from measurements – min and max [N]		Data from literature – min and max [N]			
	Jaw 1º experiment	Maxilla 2º experiment	Ricketts 1991	Proffit, 1995	Silva Filho 2007	Tanne 1994*
Lateral inferior incisors	0.22 0.23		0.20 0.44	Intrusion/ extrusion 0.10 0.59 Inclination 0.4 0.6		
Maxillary suture		16 34			10.7 44.9	9.8

* Average data.

5 CONCLUSION

The biomechanical model using the Finite Element Method showed to be an efficient tool to provide the location, distribution and analysis of applied loads. It is possible to develop a correlation between the dental movement theory, growth and bone remodeling and an adequate control of forces and pressures over living structures, leading the optimization of the treatments, reducing of pain symptoms and injuries due to the load application in the involved oral structures.

ACKNOWLEDGMENTS

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Simulation of the behavior of mandibular periimplant bone with a remodeling model

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ABSTRACT: Crestal bone loss is one the reasons for failure of dental implants and may be due, among other reasons, to mechanical loading. Bone remodeling (BR) is the physiological process by which bone adapts itself to the mechanical environment. Recently, the authors have developed a mathematical model of bone remodeling, which is used in this work to study the behavior of periimplant bone and the influence of the diameter and length of a dental implant on its long-term stability. Porosity and microstructural damage of the periimplant bone are the variables analyzed here to characterize the quality of bone. The results show that damage and porosity increase as length and especially as diameter decrease. That increase of damage and porosity is localized, as many other studies confirm, at the implant neck where a stress concentration exists.

1 INTRODUCTION

Crestal bone loss is one of the most common reasons of implant failure (Parr et al. 1988). One of the factors behind that crestal bone loss is the stress concentration at the implant neck. Bone remodeling (BR) is the process responsible for the adaptation of bone to external loads, and it involves resorption of the old bone and formation of new bone. Many mathematical models of BR can be found in the literature (Beaupré et al. 1990; Carter et al. 1989; Huiskes et al. 1987). Only a few models consider the biological aspects of this process (Hazelwood et al. 2001; Hernandez et al. 2001; Huiskes et al. 2000; Taylor et al. 2004) which are also important in BR, along with the purely mechanical ones. Recently, we have developed a mechano-biological anisotropic BR model (Martinez-Reina et al. 2009) which is used in this work to simulate the evolution of the bone surrounding a dental implant. Together with this main purpose, we have also analyzed the influence of the length and diameter of the implant. Though the effect of these parameters on the stresses are well known (Tada et al. 2003; Himmlova et al. 2004; Georgiopoulos et al. 2007; Baggi et al. 2008), our goal is to investigate the influence of these stresses on the bone density.

2 MATERIALS AND METHODS

2.1 Bone remodelling

Bone remodeling has been simulated with a mechano-biological model which follows the

algorithm schematized in Figure 1. The details of the model have been omitted for the sake of brevity, but can be consulted in (Martinez-Reina et al. 2009).

2.2 FE model: Geometry and materials

The strains and stresses in bone, needed to assess the mechanical stimulus and the accumulated damage, are calculated by means of the FEM. In this case, a model of a section of a human mandible around the first right molar has been created (see Fig. 2). This tooth has been assumed to be replaced by a prosthesis, but only the body of the implant has been included in the model. The

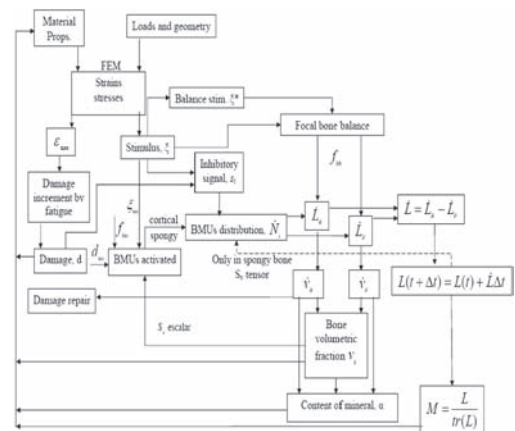


Figure 1. Algorithm of the bone remodeling model.

mechanical properties of the mandibular bone before implantation of the prosthesis have been taken from a previous work (Reina et al. 2007). The section around the first molar mentioned above is a submodel of the mandible analyzed in (Reina et al. 2007) which was cut at the boundaries CB1 and CB2 (see Fig. 2) in order to reduce the computational cost.

The bone that fills the alveolus after dental loss has a particular structure, as it can be seen in some CT scans of the original mandible (Reina et al. 2007) (see Fig. 3): an upper layer of dense trabecular bone upon a volume of a more porous tissue. So, the periimplant bone has been assigned an initial density distribution like the one shown in Figure 3, where two zones with different densities $\rho_1 = 1.5$ and $\rho_2 = 1.2$ g/cm³ have been distinguished. A simulation with a different density of the second area ($\rho_2 = 1.1$ g/cm³) has been conducted to study the importance of the initial

bone quality in the implant long-term stability. The periimplant tissue was assumed to be initially undamaged ($d = 0$).

The implant analyzed in this study has a very simple geometry: cylindrical with a semi-spherical apex, very similar to IMZ® implants. It is made of titanium with neither coating nor surface treatment. This type of implant is out of use nowadays, being the threaded ones easier to find among the commercial designs. However, most of the commercial implants are non-threaded at the neck, the area of interest of this study. In order to get good results at this area, the local geometry must be carefully modeled, while the details of the rest are not so relevant. That is the reason why such a simple geometry has been used.

The implant has been assumed to be perfectly osseointegrated all over its surface.

Two implants with different diameters (3 and 4 mm) have been analyzed to study the influence of that parameter.

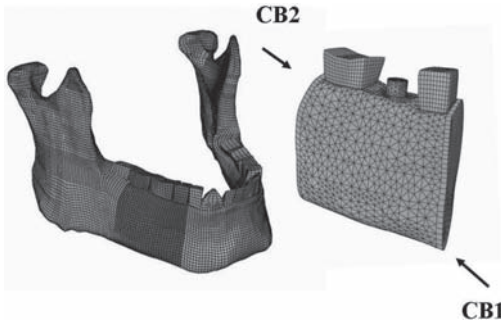


Figure 2. FE mesh of human mandible (Reina et al. 2007) (left) and a section of it (right) with a cylindrical implant replacing the first right molar.

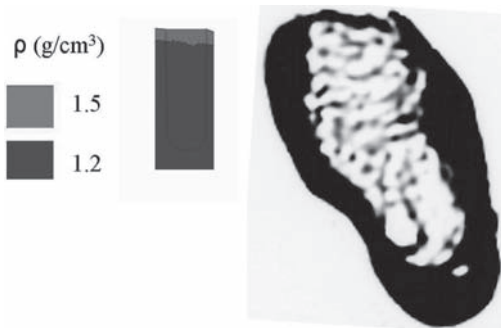


Figure 3. Computer tomography taken from the original mandible at the region of the first right molar. The original tooth had been lost and the alveolus closed by a thin layer of dense trabecular bone. So, the initial porosity distribution of the periimplant bone has been assumed as depicted to the left.

2.3 FE model: Boundary conditions

In the model of the complete mandible (Reina et al. 2007), the forces exerted by the masticatory muscles were distributed over the insertion area of each muscle. The orientation of these forces was taken from Koriath et al. (1988), and their magnitude from Nelson (1986), who provides the forces exerted by the masticatory muscles of each side during clenching and chewing with different teeth. The load cases simulated in this work are: *LM1*, mastication with the first left molar; *LM2*, with the second left molar; *RM2*, with the second right molar and *Imp*, mastication with the implanted prosthesis replacing the first right molar. In fact, this load case replaces *RM1* of the complete model (Reina et al. 2007), mastication with the first right molar.

The masticatory pattern simulated here is an alternating unilateral one, the most common as stated by Manns and Diaz (1988). So, the following sequence has been assumed: *LM1-Imp-RM2-LM2*. A total of 500 masticatory cycles have been assumed to occur every day, evenly divided among the four tasks simulated.

3 RESULTS

The numerical model provides the temporal evolution of microstructural damage, d , bone volume fraction, v_b , ash fraction, α , and fabric tensor, related with the anisotropy (Cowin, 1985) of the mandibular bone and particularly of the periimplant bone, which is of interest in this study. The first two variables, d and v_b , are studied next,

as they are fundamental to analyze the bone integrity and the stability of the implant.

3.1 Microstructural damage

Figures 4 and 5 show the distribution of d in the periimplant bone after 1000 days for both diameters in the cases $\rho_2 = 1.2 \text{ g/cm}^3$ and $\rho_2 = 1.1 \text{ g/cm}^3$ respectively.

Figures 6 and 7 show the temporal evolution of the averaged damage, $\tilde{d}$, defined as:

$$\tilde{d} = \frac{\int_{V_{peri}} (h/k) dV_T}{\int_{V_{peri}} dV_T} \quad (1)$$

in the cases $\rho_2 = 1.2 \text{ g/cm}^3$ and $\rho_2 = 1.1 \text{ g/cm}^3$ respectively. h is the damaged tissue volume per unit volume ($h = V_d/V_i$); k is a constant equal to 0.000333 and V_{peri} is the volume of bone surrounding the implant up to a distance of 2 mm approximately.

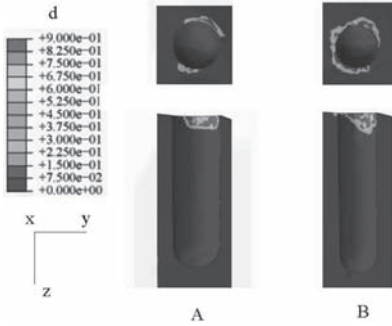


Figure 4. Distribution of microstructural damage in the periimplant bone for the two diameters after 1000 days of simulation in case $\rho_1 = 1.5 \text{ g/cm}^3$ and $\rho_2 = 1.2 \text{ g/cm}^3$.

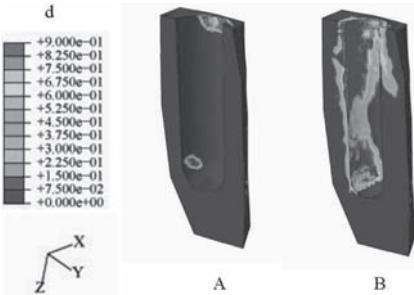


Figure 5. Distribution of microstructural damage in the periimplant bone for the two diameters after 1000 days of simulation in case $\rho_1 = 1.5 \text{ g/cm}^3$ and $\rho_2 = 1.1 \text{ g/cm}^3$.

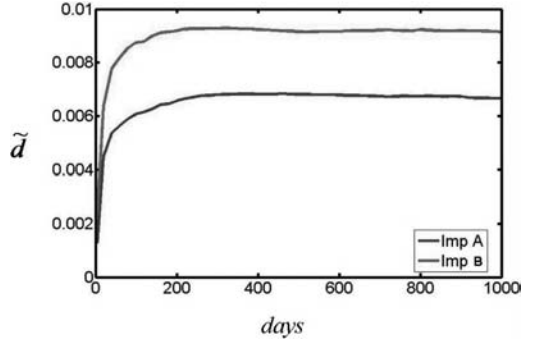


Figure 6. Temporal evolution of the averaged damage $\tilde{d}$, in the two cases studied with $\rho_2 = 1.2 \text{ g/cm}^3$.

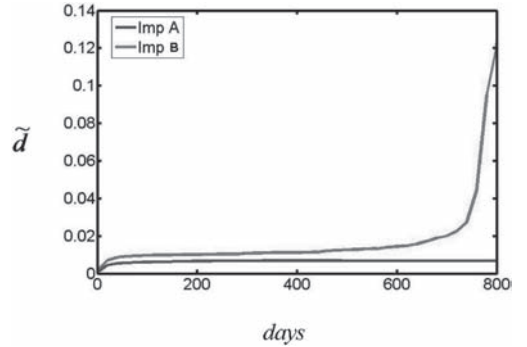


Figure 7. Temporal evolution of the averaged damage $\tilde{d}$, in the two cases studied with $\rho_2 = 1.1 \text{ g/cm}^3$.

3.2 Bone volumetric fraction

Figure 8 shows the distribution of v_b for both diameters after 1000 days, in the crestal region of the periimplant bone and in the case $\rho_2 = 1.2 \text{ g/cm}^3$.

Figure 9 shows the temporal evolution of the averaged bone volume fraction, $\tilde{v}_b$, defined in an analogous way to $\tilde{d}$ in equation (3). v_b has been divided by the initial value to facilitate the comparison between both diameters.

3.3 Discussion of results

Damage is especially high around the neck of the implant (see Fig. 4) and drastically reduced as moving away from it. Moreover, the damage level is higher for the smaller diameter. Although not as high as at the neck, stresses are still high at the apex. On the contrary, in the bone surrounding the middle part of the implant stresses are relatively low. In the case $\rho_2 = 1.1 \text{ g/cm}^3$ (Fig. 5) damage extends over a much larger surface, as bone stiffness decreases. This makes strains to be higher and

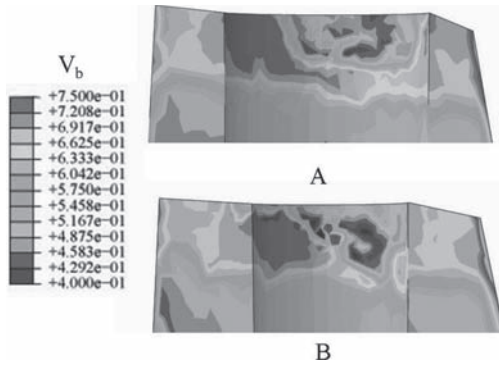


Figure 8. Distribution of v_b in the periimplant bone in the two implants after 1000 days of simulation.

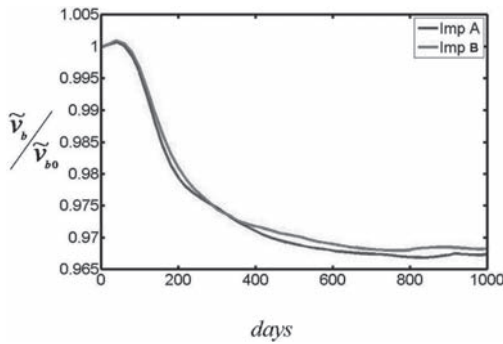


Figure 9. Temporal evolution of the averaged bone volumetric fraction in the two implants studied.

damage to accumulate in a faster rate than in case $\rho_2 = 1.2 \text{ g/cm}^3$.

The temporal evolution of $\tilde{d}$ (see Fig. 6) shows a marked initial increase of damage in every case, something expected as the initial damage was assumed to be zero. That increase of damage induces the activation of BMUs, in order to repair the damaged tissue. Finally, damage accumulation and repair balance each other out (see Figures 6 and 7), except in the narrowest implant (with a higher stress concentration at the neck) and $\rho_2 = 1.1 \text{ g/cm}^3$. In this case, damage rises very fast around day 700 leading to the implant failure.

As the formation phase is delayed with respect to the resorption phase in BMU's activity, a sudden increase in the population of BMUs (like the one observed in these simulations due to the rise of damage) produces an immediate increase of porosity, which gets stabilized after a certain time (see Figure 9). The final porosity is greater than the initial one if the population of BMUs is kept high, due to the volume of resorption cavities.

Both implants have a similar evolution of $\tilde{v}_b$, though damage is higher in implant B. This would

lead to a higher value of porosity, due to the greater population of BMUs. However, the mechanical stimulus is higher in implant B, making osteoblasts to form more tissue in this case and balancing the effect of the greater number of resorption cavities.

4 CONCLUSIONS

A recently developed model has been applied in this study in order to simulate the process of bone remodeling around the neck of a dental implant. The variables obtained here are porosity and damage, which give a good assessment of bone quality and stability of the implant.

The model has been applied here to an implant with a very simple geometry in order to study the integrity of the bone just around the neck. However, it can be applied to more complex geometries, for instance to threaded implants if a possible bone resorption around the threads is under investigation.

The results show that the bone around the neck of the implant is somewhat delicate and its integrity can be threatened by the loading conditions and the local geometry. The reason for this is the local stress concentration arising from the different Young's moduli of bone and titanium. The local high stresses and strains damage the surrounding bone and this damage induces the activation of a considerable number of BMUs whose aim is repairing that damage. The immediate effect is a rapid increase of porosity, which, along with the high damage level, might temporarily compromise the stability of the implant. The stiffness of the periimplant bone falls significantly around the neck, due to both effects leading to a short-term mobility of the implant which could cause pain and other complications. This loss of stiffness is more serious if the density of the surrounding bone is initially lower, in which case the implant might fail. In the long-term, if the implant has passed the initial complications, there is a stabilization of damage and porosity and it can be stated that the periimplant bone is in good condition.

Reducing the diameter of the implant may endanger its stability, due to the higher stress concentration around the neck that other authors have reported (Chun et al. 2002; Himmlova et al. 2004; Iplikcioglu et al. 2002; Tada et al. 2003; Tepper et al. 2002).

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Damage detection in a human premolar tooth from image processing to finite element analysis

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ABSTRACT: Digital image acquisition devices allow to quantitatively assess changes in teeth by image analysis techniques. A computerized two-step procedure was generated and applied for damage detection. In the first step, dental images were digitally processed by a segmentation method in order to identify the damage. The considered morphological properties were the enamel thickness, the number of fragments in which the enamel is chipped, and the Euler number. The information retrieved by the data processing of the section images allowed to orient the stress analysis towards selected portions of the tooth. In the second step, the stress states were analyzed exclusively in the critical zones by the finite element method, in order to numerically evaluate the risk of failure through the quantification of the von Mises stress. Coupling image processing and stress analyses turned out to be a reliable tool for damage identification in tooth mechanics.

1 INTRODUCTION

In the last few years the improvement in digital image acquisition devices had allowed some attempts to quantitatively assess changes in teeth by image analysis techniques. In Kaczmarek et al. (2003), the boundary between the enamel and the dentin was detected and quantitative measurements were performed. The quantitative results allowed to show the range of carious lesions in fissure sealant and unprotected human premolars: the aim was to quantitatively assess caries changes of teeth by digital image analysis, that was performed manually. In Imbeni et al. (2005), the propagation of cracks initiated in the enamel was studied and the results were confirmed by visual inspection considering images taken by the scanning electron microscope (SEM). In Kruzic et al. (2003), the effect of hydration over crack blunting and on the fracture mechanics are studied; microscopy and x-ray tomography were considered. Also in this case the images were considered to confirm the experimental results and no image analysis was performed. In Kantapanit et al. (2001), dental caries lesions were detected using deformable polygonal templates and edge information of teeth were found by the Canny edge detector.

The use of CT dental images has been considered in Cucchiara et al. (2004) for dental

implants. A computer vision approach was applied to identify, from the reconstructed 3D data set, the optimal cutting plane specific to each implant, to have the correct morphological measurements and to obtain the best view of the implant site.

Previous finite element (Grippio 1992, Borcic et al. 2005) and strain-gauge studies (Owens & Gallien 1995) have found that stresses concentrated in the thin cervical enamel area, and the magnitude of these stresses exceeded the known failure stresses for enamel. Improved computer and modelling techniques render the finite element method (FEM) a very reliable and accurate approach in biomechanical applications.

The aim of this paper is to outline a computerized two-step procedure for detection of damage due to non-carious cervical lesions (NCCL) in human premolars. These lesions are characterized by the loss of dental hard tissue at the cement-enamel junction (CEJ) (Rees 2002, Levitch et al. 1994). In the first step, dental images in the surroundings of the CEJ are digitally processed by the segmentation method, in order to identify the damage. In the second step, the stress state is analyzed in the critical zones—evidenced in the previous step—by the finite element method, in order to numerically evaluate the risk of failure.

2 MATERIALS

The mechanical properties (E = Young's modulus, ν = Poisson's ratio) of the materials are: $E = 80$ GPa, $\nu = 0.3$ for enamel (Rees et al. 2003), $E = 10.6$ GPa, $\nu = 0.31$ for dentine (Eskitascioglu et al. 2002), $E = 0.0021$ GPa, $\nu = 0.45$ for pulp (Lin et al. 2001), and $E = 0.0689$ GPa, $\nu = 0.45$ for periodontal ligament (Eskitascioglu et al. 2002). The materials of the various tooth structures were assumed to be isotropic, homogeneous and linearly elastic. They remained constant under the monotonically and statically applied loads.

3 METHODS

3.1 Generalities

In the following subsections a computerized two-step procedure is outlined for detection damage due to NCCL in human premolars. In the first step (subsection 3.2) dental images (Fig. 1) in the surroundings of the CEJ are digitally processed by a segmentation method, in order to identify damage. In the second step (subsection 3.3) a geometric model (Fig. 1) based on CT images of both tooth and periodontal ligament has been constructed. The height of the tooth is 23.74 mm. The stress states both in normal and malocclusion were investigated in the critical zones by the FE method in order to numerically evaluate the risk of failure.

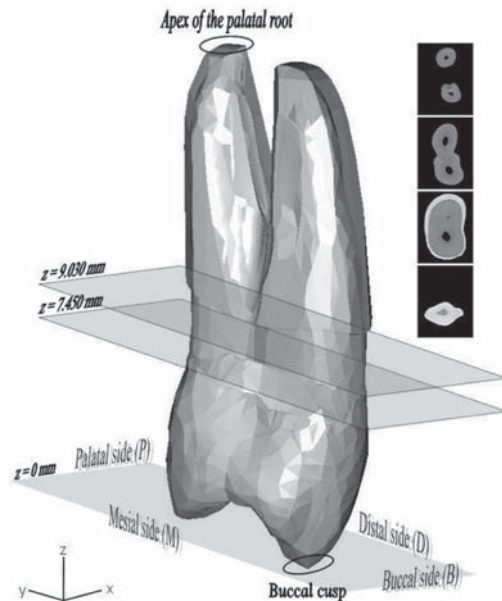


Figure 1. Model geometry and CT images of the premolar tooth.

3.2 Image processing

Digital image analysis of the complete set of the horizontal sections of the upper left first premolar was performed to check if a fracture was present and where was located. Though the images appeared of good quality a preprocessing was advisable. A γ -correction was considered to enhance the enamel. Given the gray level r , the γ -correction is the non-linear operation: r^γ . Note that for $\gamma > 1$ an image darker than the input one is obtained. Opposite effects are generated with $\gamma < 1$. Since the enamel is the region of the image with the higher gray level, for the considered sequence a gamma correction with $\gamma = 1.5$ was chosen and with this correction the background (the dentin and the pulp) is darkened and therefore the enamel has been enhanced. A greater value of γ would have enhanced also the noise. To univocally identify the enamel and distinguish the latter from the dentin a segmentation is needed. Image segmentation is a partition of the image into regions homogeneous with respect to some properties, the gray level, the texture, the color, the shape, and so on, with no loss of the information of interest. In this case, as already noted, what characterizes the enamel is its gray level that is the higher in the images. Therefore, to identify the enamel, a segmentation with respect to gray level was considered (Gonzales & Woods 2002). A n -level segmentation produces a cartoon-like image. In particular a two-levels segmentation, that is a binarization, is a white and black image. A simple method to binarize an image is to choose a threshold and assign value one to all the pixels with gray level greater than the threshold and zero to the others. The result obviously strongly depends on the chosen threshold. The Otsu method (Otsu 1979) allows to find the optimal threshold to binarize the image. It can be used also for n -levels segmentation. For the considered images a four levels segmentation was advisable. The Otsu method was applied hierarchically: first a binarization, obtaining a black and white image, and then each region (the white region and the black one) is further binarized, obtaining the four levels segmentation. The obtained segmented image is a simpler representation of the original data. The enamel is the object with the uniform higher gray level (represented in white color) and by simple logical operation it can be easily identified. On this image useful information may be retrieved, like morphological properties of the objects present in the scene, e.g. their area and geometrical characteristics.

3.3 Finite element method

In this paper, three-dimensional finite element analyses were performed on a human intact maxillary first premolar in order to address the

problems mentioned in the Introduction. An accurate finite element model based on CT images of both the tooth and periodontal ligament has been employed. Tetrahedral elements have been used to construct the model and the contact options of full bond between periodontal ligament and maxillary bone have also been used.

The finite element model was constructed from the contours of each morphological entity (dentine, enamel, periodontal ligament and pulp) obtained from successive 541 CT-images (Fig. 1). The CT-images were available at a spacing of 43.882 μm , thus allowing an accurate description of tooth anatomy to be obtained (Fig. 1).

The outline of the periodontal ligament 0.3 mm wide was generated using the outline of the tooth as a guide. The dimensions of the periodontal ligament were derived from the literature (Lindhe & Karring 1989, Schroeder & Page 1990). The solid model was transferred into the FEM program Comsol 3.4. A three-dimensional mesh was created, and the stress distribution analysis was performed. Boundary conditions have been established on the outer surface of the surrounding ligament. It has been estimated that the boundary conditions were applied far enough from force application point to not significantly influence the stress distribution in different part of the tooth. Therefore, the ligament was clamped (all displacements fixed), thus preventing rigid body displacements in directions of all three coordinate axes.

Nine-noded tetrahedral elements were applied in the discretization of the tooth morphology, resulting in 152,052 elements and 29,487 nodes with 657,543 degrees of freedom. In the case of normal occlusion, the three forces were applied on the occlusal surface. Three forces 70 N were applied orthogonally to the palatal incline of the buccal cusp, to the buccal incline of the palatal cusp, and to the palatal incline of the palatal cusp. In case of malocclusion, the force 200 N was applied normally to the buccal incline of the palatal cusp (Nakamura et al. 2001, Borcic et al. 2005).

4 RESULTS

4.1 Data processing

A sequence of 50 images was considered. Each image is a matrix of dimension $n \times m$ pixels, with $n = 233$, number of rows, $m = 179$, number of columns. The pixel length is equal to 0.0439 mm. For each cross section of the considered sequence of tooth images the γ -correction and the four levels segmentation were performed. In Figure 2 the cross section at level $z = 8.38$ mm is analyzed: in Figure 2a and Figure 2b the original picture and its γ -correction with $\gamma = 1.5$ are displayed respectively.

In Figure 2c the four level segmentation of Figure 2a obtained by the hierarchical Otsu method is shown. Once the image has been segmented, the enamel is univocally identified by simple logic operation as the region with higher gray level (Fig. 2d). For each available tooth cross section, the enamel may be characterized by its area, its thickness in distal (D), mesial (M), buccal (B) and palatal (P) sides, the number of regions that constitute the enamel itself and the Euler number of the enamel. The area is the number of pixels constituting the enamel. The thickness is evaluated along the above mentioned zones over a grid superimposed to the image cross section. A grid spacing equal to $n/4$ along the buccal-palatal (B-P) direction and equal to $m/4$ along the mesial-distal (M-D) direction was considered, see Figure 3. The chosen grid intersects the identified enamel in twelve segments (three for each side) whose lengths, representing the thickness of the enamel in these sections, have been automatically evaluated. The Euler number is the difference between the number of object and

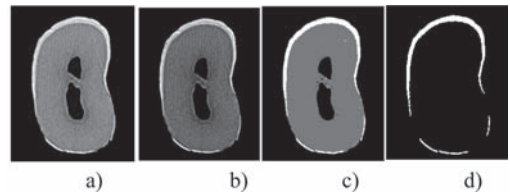


Figure 2. Identification of the enamel from cross section $z = 8.38$ mm: a) original image; b) gamma correction; c) four levels segmentation; d) enamel identification.

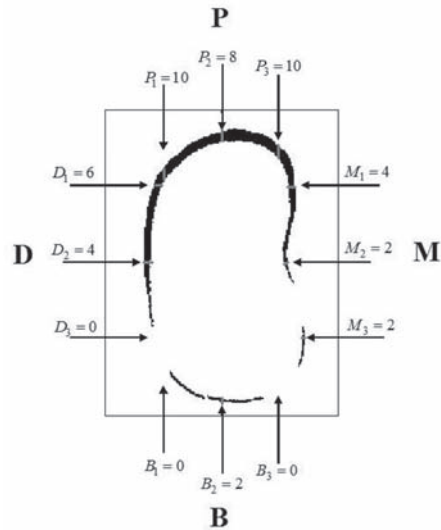


Figure 3. Thickness values along the distal (D), mesial (M), buccal (B) and palatal (P) sides.

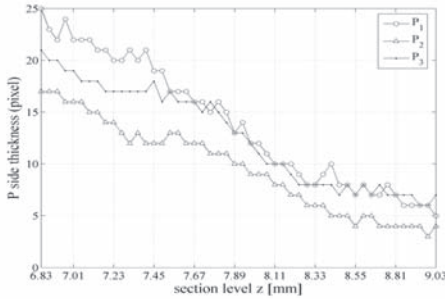
the holes present in the image. When the enamel constitutes a unique not fragmented object (similar to an elliptic ring) the Euler number is equal to zero (one object, the enamel, minus one hole, that is the space inside the enamel).

For example, for the image of Figure 2d the larger fragment has area equal to 1614 pixels, whereas the others have area equal to 79, 85 and 47 pixels; the number of regions, equal to four, that compose the enamel has been automatically evaluated along with the Euler number, also equal to four.

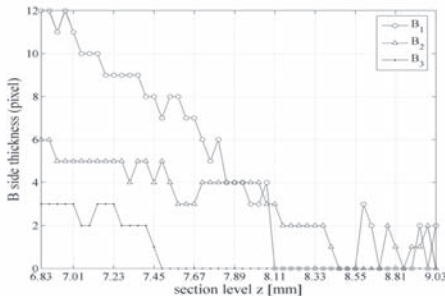
This information helps in determining whether the enamel presents one or more lesions. More precisely, analyzing the sequence of cross sections, a sudden change in one of the considered properties may indicate the presence of a lesion. In some cross sections the enamel is fragmented. It is useful to count the number of pieces in which the enamel is divided and the Euler number and where the fracture took place. All these data are collected for each cross section of the available sequence.

In Figure 4a and Figure 4b the thickness (P_1 , P_2 , P_3) and (B_1 , B_2 , B_3) relative to the buccal-palatal direction, palatal side (B-P/P) and relative to the buccal-palatal direction, buccal side (B-P/B), evaluated in the segments intersected by the chosen grid on the enamel are represented respectively.

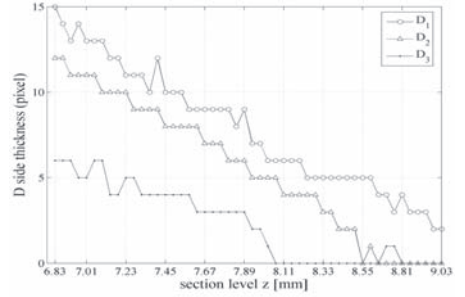
Analogously, in Figure 4c and Figure 4d the thickness (D_1 , D_2 , D_3) and (M_1 , M_2 , M_3) relative to the distal-mesial direction, distal side (D-M/D) and



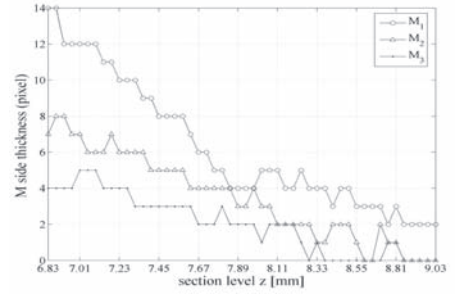
a)



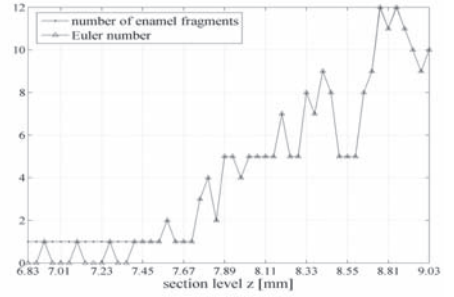
b)



c)



d)



e)

Figure 4. Analysis of the sequence of cross sections corresponding to $z = 6.83 + 9.03$ mm: a)–d) thickness in the B-P/P, B-P/B, D-M/D D-M/M sides: each of them has been evaluated in three points, according to the positions indicated in Figure 3; e) number of enamel fragments with Euler number for each cross section.

relative to the distal-mesial direction, mesial side (D-M/M), evaluated in the segments intersected by the chosen grid on the enamel are represented respectively. In Figure 4e the number of enamel fragments and the Euler number for each cross section are represented.

4.2 Stress analyses

A detailed description of the stress distribution was based on 30 horizontal sections at the CEJ (Fig. 1). In particular, Figure 1 reports the levels

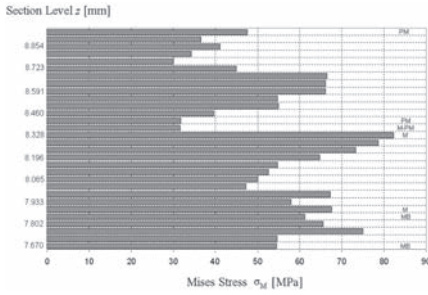


Figure 5. Histogram of the Mises Stress σ_M [MPa] in normal occlusion.

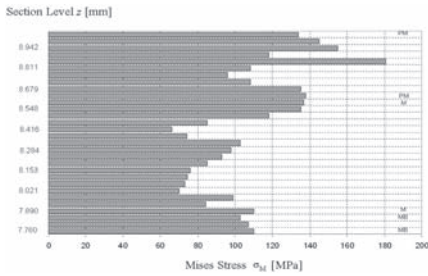


Figure 6. Histogram of the Mises Stress σ_M [MPa] in malocclusion.

$z_1 = 7.45$ mm and $z_2 = 9.030$ mm of the extremal cross-sections measured with respect to the buccal cusp. Figure 5 and Figure 6 show differences in the stress distribution between the model under different loading conditions. Two typical cases have been considered of normal occlusion (Fig. 5) and malocclusion (Fig. 6). Mises stress (σ_M) field was analysed to estimate the risk of failure in the models. In more detail, each row of the histograms in Figs. 5–6 refers to a single cross-section and indicates the sector and the tissue where the maximum value is attained by the stresses σ_M at the relevant section. Each cross-section is identified by its level $z = 7.670 \div 9.030$ mm (Fig. 1) on the ordinate axis.

5 DISCUSSION

5.1 Data processing

The Otsu segmentation method enjoys good performance with numerical efficiency. This is the reason for which it has been preferred to other segmentation methods. The thickness in the P side decreases showing no irregular behavior; in fact in this direction no fracture is present.

From level $z = 7.67$ mm it can be noted that the thickness in the M and B sides is less than 6 pixels (corresponding to 0.2634 mm), as shown in

Figure 4b, where B_3 thickness is equal to zero, and Figure 4d. The enamel is constituted by one fragment and the number of Euler is equal to one: these two characteristics mean that a fracture is present.

From level $z = 8.63$ mm the damaged zone begins and in the D and M sides the thickness becomes less than four pixels (corresponding to 0.175 mm).

A global overview of all the collected information allows to assert that in the lower part of the M side and in the B side, a weakness in the enamel is present. This becomes significant, thickness less than 0.0878 mm in the B side and less than 0.1756 mm in the M side, at level $z = 8.328$ mm ($\sigma_M = 82.3$ MPa in normal occlusion), and at level $z = 8.854$ mm ($\sigma_M = 181$ MPa in malocclusion), with thickness less than 0.0439 mm in the B side and less than 0.1756 mm in the M side. In Figure 7 the segmented images of the above levels are shown respectively.

The detailed screening carried out in this subsection on the section images of the tooth at various levels allowed to select the most critical zones and hence to conduct specific analyses of the stress states thereby. This investigation is demanded to the next subsection. In particular, the attention was devoted to the buccal, mesial and palatal-mesial sectors, where the thickness of the enamel layer underwent a dramatic drop. The stress analyses were performed within the level range $z = 7.670 \div 9.030$ mm and were extended to loading conditions of both normal and malocclusion.

5.2 FEM analyses

Figure 5 shows that in the range $z = 7.670 \div 7.845$ mm the local maximum value ($z = 7.758$ mm, $\sigma_M = 75.0$ MPa) of the Mises stress in normal occlusion is located in the MB sector on the outer surface of the enamel. From $z = 8.196$ mm to $z = 8.942$ mm the highly stressed zones migrate toward the PM sector, passing through the M-PM sector ($z = 8.371$) and the M sector (from $z = 7.889$ mm to $z = 8.328$ mm). They concentrate where the enamel gets thinner and vanishes leaving place to

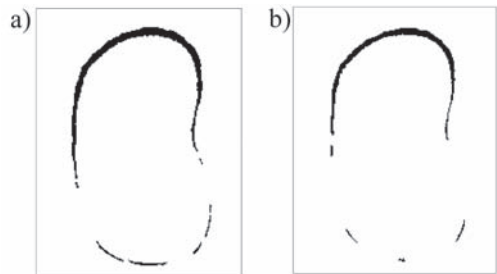


Figure 7. Enamel identification for the cross section at level $z = 8.328$ mm (a) and $z = 8.854$ mm (b).

the dentin. The absolute maximum is reached in the PM sector at $z = 8.328$ mm and is 82.3 MPa.

With reference to Figure 6, the largest values of the Mises stress in malocclusion are uniformly distributed in the MB ($z = 7.760 \div 7.845$ mm) and M ($z = 7.889 \div 8.591$ mm) sectors of the enamel near the interface with the dentin; then, they tend to concentrate in the PM sector ($z = 8.635 \div 9.030$ mm) on the outer surface of the enamel, where they reach the maximum value 181.0 MPa at $z = 8.854$ mm. For completeness's sake, the other two local maxima 110.5 MPa at $z = 7.760$ mm and 137.3 MPa at $z = 8.591$ mm are attained respectively in the MB sector ($z = 7.760 \div 7.845$ mm), and in the M sector ($z = 7.889 \div 8.591$ mm).

It should be remarked that both absolute and local maxima of the Mises stress were attained exactly at the sections where the data processing performed in Subsects. 4.1 and 5.1 found out minimal thicknesses of the enamel layer.

6 CONCLUSIONS

In this paper the analyses of CT image sequence provided information about variations of the thickness, area and number of enamel fragments in a premolar tooth by a segmentation procedure. This investigation has been considered not only to confirm experimental results or to verify *a posteriori* lesions on the tooth, as is currently done in literature but to *a priori* localize damage via automatic processing. The analysis of the morphological properties, namely enamel thickness in different points and the number of fragments in which the enamel is divided, along with the Euler number, allowed the identification of the zones of the enamel in which a weakness is present.

The information given by the above-described data processing led to the successive FEM analyses of the constructed model of the premolar geometry, in order to limit the mechanical characterization exclusively to the critical zones. In fact, significant jumps were predicted of the von Mises stress values in the enamel at the cervical area in two different loading conditions of normal and malocclusion. As expected, noticeably weakening in the continuity of the structure of the hard dental tissues causes the increase of the stresses in the cervical region.

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Fracture behaviour of some heat curing dental resins

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ABSTRACT: Mechanical behavior of complete dentures is a priority among experimental papers, in order to define the fracture mechanism of acrylic resins. Paper presents experimental results concerning the manner of fracture of some heat curing materials from acrylic resins class, respectively, MELIODENT, SUPERACRYL, TRIPLEX and VERTEX. The fracture samples were examined on stereomicroscope Olympus. The difference in mechanical behavior is explained in terms of stereomicroscopic analysis, in longitudinal and transversal cross sections of the tested samples. Finally, some conclusions concerning the mechanism of fracture were drawn in correlation with the aspect, shape, dimension and proportion of reinforced small chop red fibers in resin matrix.

1 INTRODUCTION

A real development in dental techniques for repairing oral teeth was permanently based on intensive utilization of different materials. Starting from existing materials in nature as bone, wood, metals and than alloys (gold, silver, stainless steel, cobalt-chromium, titanium), nowadays, based on intensive techniques, some new and performed materials may be used, as polymers (rubber, resins), up to composite materials based on acrylic resins reinforced with fibers. In XIX-th century plastic materials and resins became favorites. Today, the acrylic resins have acceptable properties, simple technology, but also disadvantages- such as dimension changes during elaboration, porosities, reduced thermal conductivity, poor mechanical resistance and low biological properties on used resins such as: diacrylic, polyacetil (polymethaethylene). Utilization of these resins is based on their ideal properties: esthetic properties (translucidity, color similar to the replaced tissues and color maintaining both during elaboration and inside human mouth) and physical- chemical properties (dimension stability and shape maintaining during processing and inside human mouth, good elasticity, wear resistance in human mouth, impermeability for saliva and food, good polishing and hygiene, good

connection to the other prosthesis components, superior plasticity temperature than human mouth temperature, non toxicity and non-irritating for the human tissue). According to EN ISO 1567 there are four types of resins: heat curing resins (upper 65°C) either bi-component, or mono-components, self curing (lower 65°C), thermoplastic materials in granulate form, light curing resins and microwave polymerized resins. Heat curing resins are now the most used materials for realization of partial or total dental prostheses. World market of bi-component heat curing resins know a lot of product, such as Meliodent (Heraeus Kulzer), Vertex (Vertex), Superacryl (Spofa), Triplex (Ivoclar). One of the most difficult prosthetic restorations is that of total dentition, where the result is immediately “to be seen.” A bad prosthesis cannot be kept on the position and isn’t stable from the first moment. Prosthetic restoration of edentulous patients must follow the recovery of dental arcades, functional almost partial mastication recovery, and why not aesthetic shape and form of teeth, occlusal equilibration, homeostasis of the prosthetic field. The complete denture is a prosthesis inserted in human mouth, with some problems in patients’ accommodation. Being mobile from oral cavity, it is sometimes embarrassing, creating some physical handicap and infirmity.

Frequently, there are non corresponding situations, when complete dentures may have cracks, producing partial or total breakdown. The problem of mechanical behavior of acrylic resins materials is a real one, being well and repeatedly presented in literature [1,2,3,4,5]. The necessity of a minimum level of mechanical characteristic values guarantee may appear in order to assure the prosthesis viability on long term. The aim of presented paper is the necessity of processes grounding which takes place for integration maintaining of bicomponent heat curing resins, and also to structural characterization of two types of resins, with differential behavior.

2 MATERIALS AND EXPERIMENTAL PROCEDURE

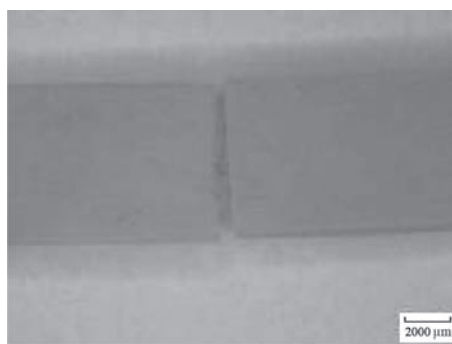
There were analysed four acrylic resins which are usually used in dental practice, respectively: MELIODENT, SUPERACRYL, TRIPLEX and VERTEX. In accordance with dental practice, there were made by polymerization samples with the following dimensions: 2 mm thickness, 30 mm length and 5 mm width. In order to define the mechanism of fracture, all amples were loaded in the same manner. Mechanical characteristics were determined by Zwick Roel equipment with data processing using testXpert system. The loaded stress was 50 kN and the rezolution 0,1 μm for all experimental samples. The experimental samples were then analysed at stereomicroscope Olympus type SZX7, equiped with image processing soft QuickphotoMicro 2.2. There were analysed both longitudinal and transversal surfaces perpendicular to fracture surface.

3 EXPERIMENTAL RESULTS AND INTERPRETATION

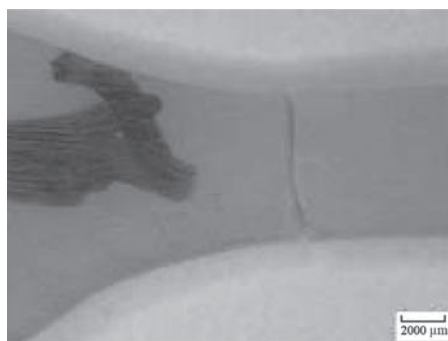
Detailed analysis under the stereomicroscope revealed interesting observations. As a general remark, the entire sample has a brittle fracture, with a quasi crystalline aspect in transversal cross section. Each resin has its own structural characteristics.

Resin type MELIODENT, given in Figure 1, has in general a brittle aspect. One may remark in transversal cross section relative many chop fibers, about 3–4 red and short fibers. All the fibers cracked separately than the matrix, as a sign of a different fracture behavior between fiber and the matrix. There is no fragmented fracture of the samples.

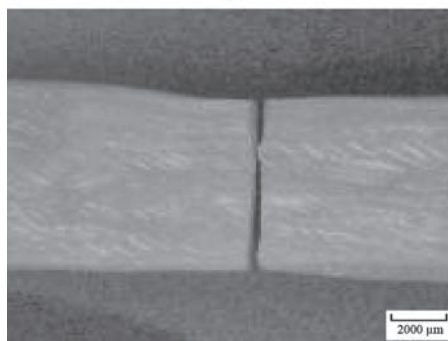
Resin type SUPERACRYL, given in Figure 2, has a specific structural feature. The entire sample



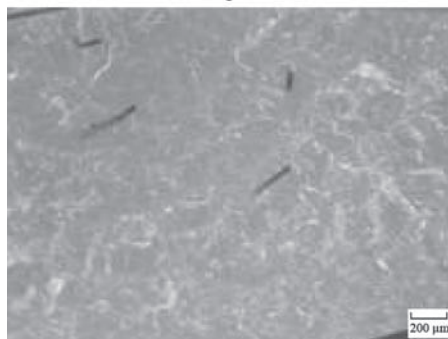
a



b



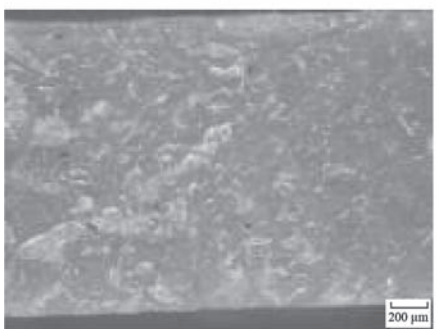
c



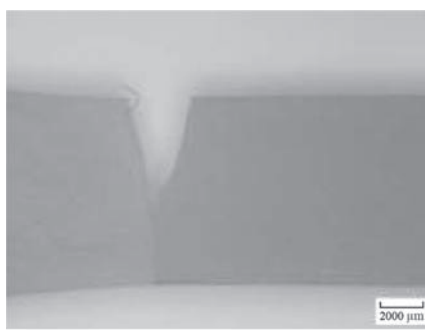
d



e



f

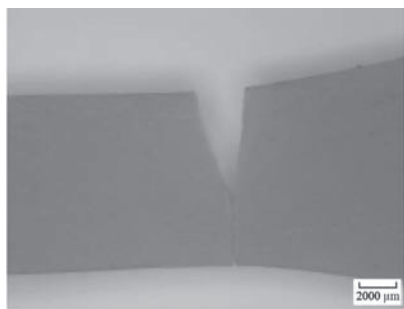


c

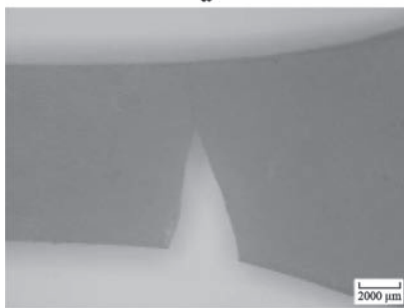


d

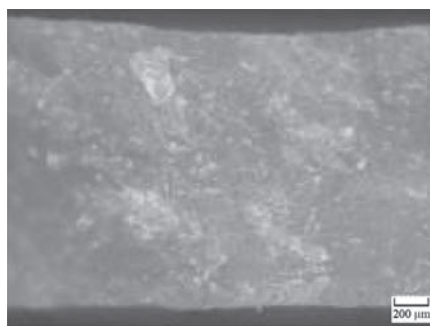
Figure 1. Stereomicrostructural aspect of MELI-ODENT dental resin: longitudinal section (a, b, c) and transversal cross section (d, e, f).



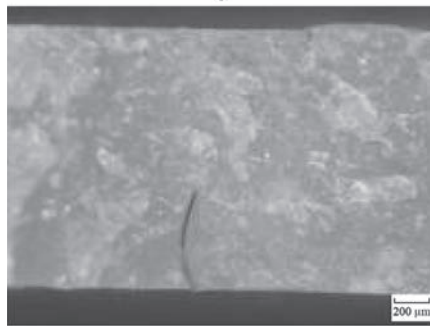
a



b



e



f

Figure 2. Stereomicrostructural aspect of SUPER-ACRYL dental resin: longitudinal section (a, b, c) and transversal cross section (d, e, f).

fractured in fragmented manner, resulting three parts after fracture (the so called 100% “fragmented fracture”). One may remark a small amount of short chop fibers on the transversal cross section, respectively one- two.

Resin type TRIPLEX, given in Figure 3, has a proportion of fragmented fracture about 70%, in three pieces. Only 20% from the total number of fibers may crack in the same time with the matrix. This resin has the highest amount of reinforced fibers among all investigated resins.

Resins type VERTEX, given in Figure 4, have also a fragmentary fracture, but in proportion of about 30%. In all situations, the fibers do not fracture in the same time with the matrix. These resins have the longest fibers among the investigated resins.

The mechanical characteristics of the heat curing dental resins are given in Figure 5. As one may remark, the lowest value of the ultimate strength is for MELIODENT, than SUPERACRYL and TRIPLEX, the highest being for VERTEX. Considering the manner of fracture we may conclude that resin VERTEX has the best mechanical behavior among all the investigated heat curing dental resins.

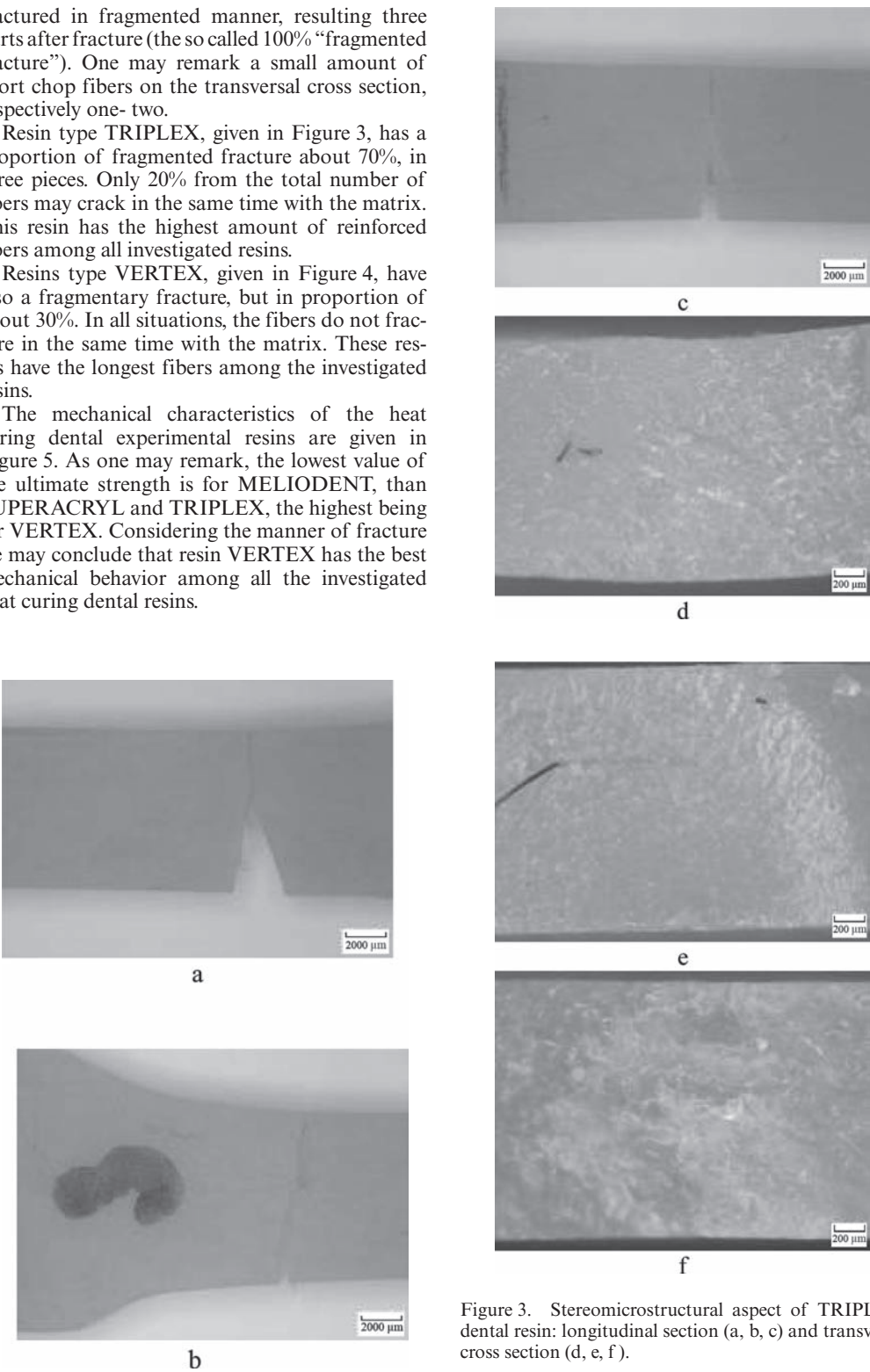
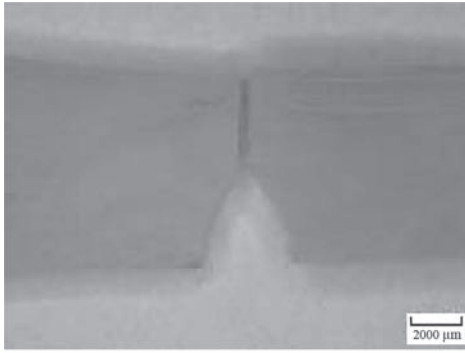
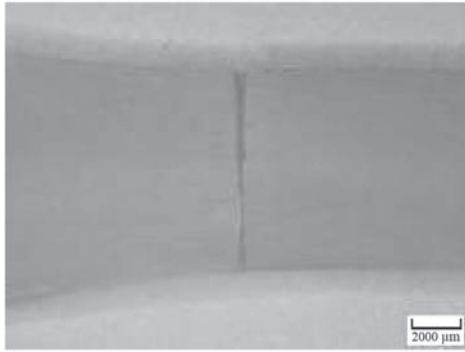


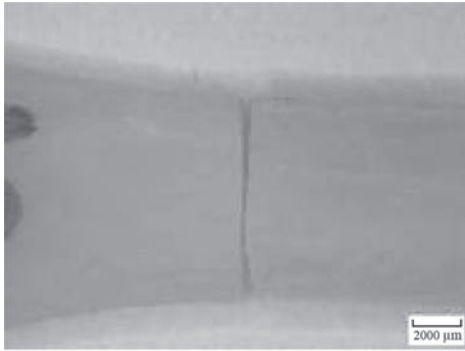
Figure 3. Stereomicrostructural aspect of TRIPLEX-dental resin: longitudinal section (a, b, c) and transversal cross section (d, e, f).



a



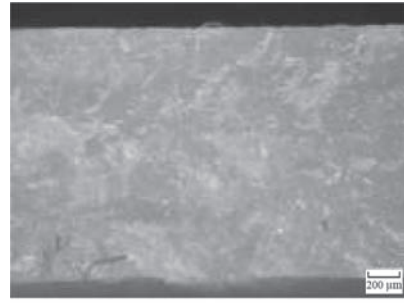
b



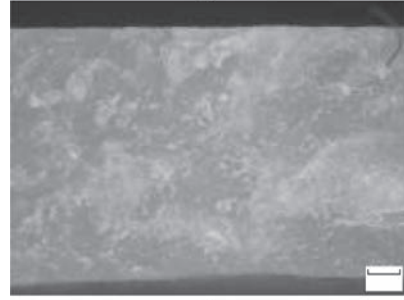
c



d



e



f

Figure 4. Stereomicrostructural aspect of VERTEX dental resin: longitudinal section (a, b, c) and transversal cross section (d, e, f).

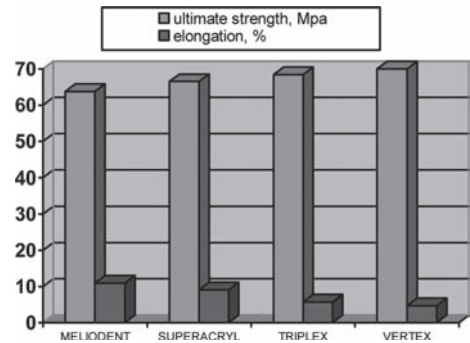


Figure 5. Mechanical characteristics of the heat curing dental experimental resins.

4 CONCLUSIONS

Comparative analysis of fracture behaviour for four heat curing dental acrylic resins, respectively type MELIODENT, SUPERACRYL, TRIPLEX and VERTEX, may reveal the following aspects:

- All the heat curing dental acrylic resins present fragmented fracture, but in different proportion: 100% for MELIODENT and SUPERACRYL, 70% for TRIPLEX and 30% for VERTEX.

- Mechanical characteristic values of samples were in different ranges: ultimate strength is 63 MPa for Meliodent, 66 MPa for SUPERACRYL, 69 MPa for TRIPLEX, the highest value being for VERTEX, respectively 70MPa.
- Different behavior of dental acrylic heat curing resins is due to significant differences between mechanical characteristics of reinforced fibers and matrix. Red small chop fibers may or may not fracture in the same time with the matrix. So, at MELIODENT, VERTEX and SUPER-ACRYL samples fibers may not fracture in the same moment with matrix, in opposite with TRIPLEX samples, where only 20% from the total number of fibers are broken simultaneously with the matrix.
- The best mechanical behavior of the investigated heat curing dental acrylic resin is for VERTEX, with the highest amount of reinforced fibers among the other resins.

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Effect of the material of the prefabricated post on restored premolars

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ABSTRACT: In a previous work, the authors studied the effect of post material on the biomechanical performance (fracture strength and stress distribution) of restored teeth without the addition of the final crown restoration. Teeth restored with stainless steel posts showed significantly lower failure loads than teeth restored with glass fiber posts. Moreover, the failure mode for glass fiber post systems allowed further repair. The aim of this work was to study how the material of the prefabricated post affects the biomechanical performance of teeth restored with the crown. Despite the protective effect of the crown, the safety factor obtained for restorations with glass fiber posts is significantly higher than that obtained for restorations with stainless steel posts.

1 INTRODUCTION

When an aesthetic appearance is required of a tooth restoration, traditionally a porcelain-fused-to-metal crown has been used. However, because of concerns about allergic reactions or biocompatibility (Sjogren et al., 2000), patients and dentists alike have come to prefer metal-free restorations. The use of ceramics makes it possible to replicate the aesthetic characteristics and vitality of natural teeth (Ku et al., 2002) and ceramic materials are superior in terms of permeability to light and biocompatibility, but they can be inherently brittle and weak when placed under tensile and torsional stresses (Hwang & Yang, 2001). The use of reinforced glass-ceramic materials lessens the impact of this drawback. Survival rates of glass-ceramic crowns, even on posterior teeth, have far exceeded those of traditional all-porcelain crowns, despite having nearly equivalent strengths and thicknesses (Fradeani & Redemagni, 2002).

In a previous work, the authors studied the effect of post material (Barjau-Escribano et al., 2006) on the biomechanical performance of restored teeth without the addition of the final

crown restoration. Significantly lower failure loads for teeth restored with stainless steel posts were found experimentally. Moreover, the failure mode for glass fiber post systems allowed further repair. The experimental results were used to validate a finite element model of a restored tooth that was developed by the authors. It was concluded from the model estimations that post systems in which the elastic modulus of the post is similar to that of dentin and core offer a better biomechanical performance.

The aim of this study was to ascertain whether those results are pertinent when considering crowned teeth, particularly maxillary premolars. In other words, is biomechanical performance (fracture strength and stress distribution) of crowned teeth better when using glass fiber posts than when using stainless steel posts? An experimental fracture strength test was performed on extracted human teeth which were restored using two different post materials (glass fiber and stainless steel) and a glass-ceramic crown. A 3D finite element model of the restored tooth was then used to analyze the stresses that were originated with the different post materials.

2 MATERIALS AND METHODS

A combined theoretical and experimental method was used, similar to that used in the authors' previous work carried out to study the influence of post material on restored teeth (Barjau-Escribano et al., 2006).

The ParaPost Fiber White and the ParaPost Stainless Steel (Coltène/Whaledent Inc, Mahwah, NJ, USA) were used for the study. These posts were selected because their geometry is similar and they are manufactured in the same sizes, but have significantly different elastic moduli (20 GPa for the ParaPost Fiber White and 207 GPa for the Parapost Stainless Steel).

2.1 Fracture strength test

First, an experimental fracture strength test was performed on endodontically treated and restored teeth. Eighteen sound human premolars, extracted for periodontal reasons, were selected for the study. Dental plaque, calculus, and periodontal tissues were removed. Patients were informed that their extracted teeth were destined for experimental purposes and written informed consent was consequently obtained from each of them. Nine specimens were restored using glass fiber posts and nine with stainless steel posts.

As fracture load has been proven not to be clearly dependent on post length (Rodríguez-Cervantes et al., 2007), the widespread recommendation (Smith & Schuman, 1998) that post length should be about three quarters of root length was used for all the specimens. The extended recommendation (Nergiz et al., 2002; Stern & Hirshfeld, 1973) that post width should not be greater than one-third of the root width, at its narrowest section, was followed.

All specimens were prepared following the same endodontic treatment described in a previous work (Barjau-Escribano et al., 2006). After endodontic treatment, teeth were embedded in individual blocks of the autopolymerizing acrylic-resin Degacryl (Degussa AG, Düsseldorf, Germany) using a mould; 1.5 mm of the root was left submerged to simulate the real conditions of insertion of teeth into the bone. The posts were introduced and cemented with the resin cement ParaPost Cement (Coltène/Whaledent Inc, Mahwah, NJ, USA) in compliance with the manufacturer's instructions. The subsequent restoration of the core was performed with the dual cure resin ParaCore (Coltène/Whaledent Inc, Mahwah, NJ, USA), also in compliance with the manufacturer's instructions. Twenty-four hours later, the core was finished with a high-speed diamond bur in order to carry out the final crown restoration. The material of the crown

was IPS Empress® (Ivoclar Vivadent AG, Schaan, Liechtenstein), which is a leucite-reinforced glass-ceramic. Cementation of the crowns was performed using Dual cement (Ivoclar Vivadent AG, Schaan, Liechtenstein). The cement was dispensed inside the crown, placed on the prepared tooth, and excess cement was removed with a micro-brush. Finally, a 40-second polymerization was performed on each side until complete hardening was achieved.

An IBERTEST ELIB-30/W (Ibertest, Madrid, Spain) universal testing machine was used for the experiment. The specimens were placed in a retention device and mounted on the universal testing machine. This device allowed the teeth to be loaded on the palatal side at a 30° angle to the radicular axis, in the vestibular direction (Figure 1), thus simulating the real direction of loads during biting (Assif et al., 1989; Cohen et al., 1997; Fokkinga et al., 2005). A controlled loading force was applied to the teeth at a rate of 5 N/s, until failure. The loading force required to cause failure was recorded, and the results for each post system were compared by means of a one-way ANOVA (SPSS 17.0, SPSS Inc, Chicago, IL, USA), with a 5% significance level.

2.2 Finite element model

The model used in this study was based on the geometry of a real tooth, obtained through a 3D scanner, and the 3D modeling software Pro/Engineer (PTC, Needham, MA, USA) was used to generate and later assemble the geometries for all the components. This model has been properly validated in previous works (Barjau-Escribano et al., 2006; Rodríguez-Cervantes et al., 2007). Figure 2 shows a longitudinal section of the geometrical model considered, including all the components that were modeled: bone (cortical and

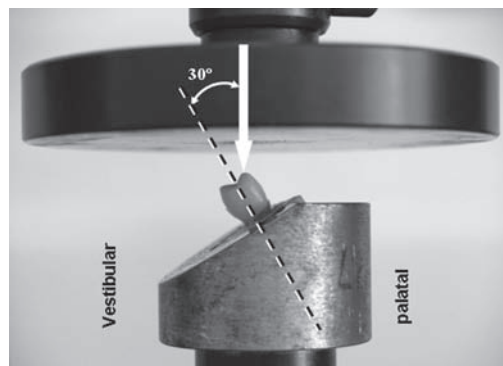


Figure 1. Experimental fracture strength test performed on teeth.

trabecular components), periodontal ligament, root/dentin, gutta-percha, post, crown and post cements, core and crown. The mechanical properties of the different components of the model were obtained from the literature and from the manufacturer of the post (Coltène/Whaledent Inc, Mahwah, NJ, USA) and from the manufacturer of the glass-ceramic crown (Ivoclar Vivadent AG, Schaan, Liechtenstein). The aforementioned properties are presented in Table 1.

The Pro/Mechanica module included as part of Pro/Engineer was used to divide (mesh) the CAD geometry. Solid tetrahedral elements with a size of 0.3 mm were used for the mesh on all the components, except on trabecular and cortical bone, where a size of 1 mm was applied. The model had 398 726 elements defined by 69 044 nodes. As boundary conditions, the displacements of all nodes on the lateral surface and base of the component representing the bone were constrained.

The analysis was carried out using the finite element analysis software MSC-PATRAN-NASTRAN (MSC.Software Corporation, Santa Ana, CA, USA). In order to study the influence of the post material on restored teeth with the addition

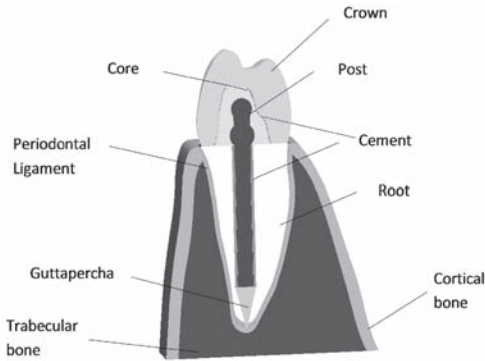


Figure 2. Section of the geometrical model that was generated. Modeled components.

Table 1. Mechanical properties of the materials used.

Component/material	E (GPa)	ν
Root/dentin	18.6	0.31
Gutta-percha	0.00069	0.45
Periodontal ligament	0.0689	0.45
Cortical bone	13.7	0.30
Trabecular bone	1.37	0.30
Post cement	5	0.30
Core	20	0.30
Crown	62	0.30
Crown cement	10	0.30
Stainless steel post	207	0.30
Glass fiber post	20	0.30

of the final crown, the study was undertaken with a model of a maxillary premolar restored using the ParaPost Fiber White and the ParaPost Stainless Steel, and the addition of the final crown. A 300 N load was applied to the palatal side of the tooth at an angle of 30° to the radicular axis in the vestibular direction, in order to simulate real biting force (as in the fracture strength test). Under this load, the stress distribution pattern of the restored tooth was studied. The stress distribution pattern provided information about the fracture mechanism of the restored tooth: for the same external load, higher stresses indicated a higher probability of reaching the failure load.

3 RESULTS

3.1 Results from the experiment

From the experimental fracture strength test, significant differences ($p = 0.019 < 0.05$) were observed between the failure loads of the different groups (glass fiber, stainless steel) of restored teeth. A box-whisker graph showing the spread of data groups around their medians can be observed in Figure 3.

3.2 Results from the model

In this section, the results of the study of the influence of the post material are presented. The von Mises stress distributions over the central longitudinal section of the tooth estimated by the model when varying post material are represented in Figure 4, using a color scale (warmer

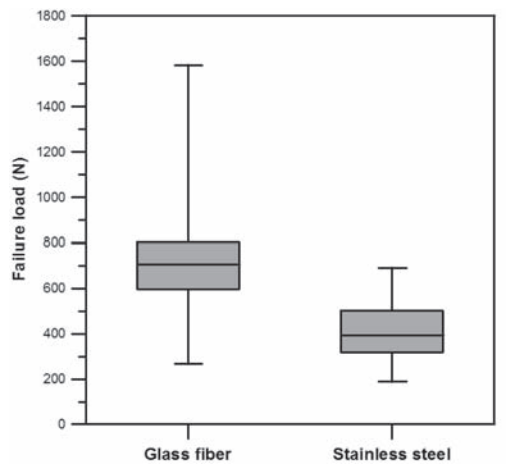


Figure 3. Box-whisker graph showing the effect of post material factor on failure loads for the groups of specimens studied.

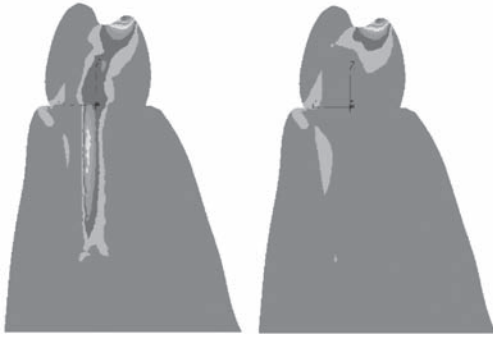


Figure 4. Von Mises stress distribution pattern (color scale: 0–120 MPa) estimated by the model using a stainless steel post (left image) and a glass fiber post (right image) for restoring teeth, on the sagittal section (left vestibular, right palatal).

Table 2. Maximal von Mises stresses estimated in the components of the model of restored teeth for the different post materials considered.

	Von Mises stresses (MPa)	
	Stainless steel post	Glass fiber post
Post	106.2	15.0
Core	36.8	15.0
Crown cement	23.1	26.4
Dentin	26.1	30.5
Crown	99.1	99.1
Post cement	84.8	16.2

colors represent higher stresses). In the case of the stainless steel post, a stress concentration was observed along the juncture of the post with dentin, cement and core, together with a stress concentration at the force application zone. In the case of glass fiber posts, no stress concentration was observed along the juncture of the post with dentin, cement and core. The maximum stresses were considerably smaller than those predicted when using stainless steel posts and were located at the force application zone. This contrasts with the results of previous works (Barjau-Escribano et al., 2006; Rodríguez-Cervantes et al., 2007), which were obtained without the final crown. In those cases, maximum stresses were located in the vestibular region, where the tooth is embedded into the bone, within the dentin and composite core.

The maximum von Mises stresses estimated at each component using the model when varying the post material are presented in Table 2 below.

Flexural strengths of each component material, obtained from the literature, are presented in

Table 3. Flexural strengths for the different component materials of the model.

	Flexural strength (MPa)	Source
ParaPost fiber white	990	Manufacturer
ParaPost stainless steel	1436	Manufacturer
ParaCore	90	(Dodiuk-Kenig et al., 2004)
Dual cement	45.1	(Saskalauskaite et al., 2008)
Dentin	212.9	(Plotino et al., 2007)
Crown	160	Manufacturer
ParaPost cement	97	(Baudin II & Burgess, 2002)

Table 3. To compare the maximal stresses among components, a safety factor can be calculated for each component as the ratio between the flexural strength of its material and the maximal von Mises stresses in the component (Tables 2 and 3). A smaller safety factor means that the component is more susceptible to failure. The model of teeth restored using stainless steel post predicts that the post cement component is the one with the smallest safety factor, followed very closely by the crown and crown cement. Taking into account the accuracy of the flexural strengths, this means that the fracture might begin either at the interface between post and cement, as occurred in the previous works without the addition of the final crown (Barjau-Escribano et al., 2006; Rodríguez-Cervantes et al., 2007) or at the force application zone.

The model of teeth restored using glass fiber posts predicts that the crown and crown cement components are the ones with the smallest safety factors and, consequently, the fracture would begin in these components, i.e., at the force application zone.

Finally, the smallest safety factor of the components can be considered as being that of the overall system. On comparing both systems, the safety factor obtained for glass fiber post restorations is significantly higher than that obtained for stainless steel post restorations, which means that the model predicts a higher fracture load for restorations using glass fiber posts.

4 DISCUSSION AND CONCLUSIONS

Within the limitations of this work, we can state that teeth restored with glass fiber posts have a considerably higher strength than those restored using

stainless steel posts. Crowned and non-crowned teeth present a different mode of failure. In the case of teeth restored with stainless steel posts, the failure occurred along the juncture of the post with dentin and core when no crown was considered (Barjau-Escribano et al., 2006; Rodríguez-Cervantes et al., 2007). When the final crown is added, the analysis of the safety factor revealed that the probabilities of failure along the juncture of the post with dentin and core and in the crown are similar. Thus, stress concentration at the post-dentin-core interface becomes less important because its influence is masked.

In the case of teeth restored with glass fiber posts, when no crown was considered, failure occurred in the vestibular region where the tooth is embedded in the bone, because of the flexion of the core (Barjau-Escribano et al., 2006; Rodríguez-Cervantes et al., 2007). When the final crown was added, the analysis of the safety factor revealed that the failure occurred at the crown, in the area where the load was applied. In any case, the mode of failure predicted by the model for crowned restored teeth would not affect the root either.

Better biomechanical performance was observed experimentally for teeth restored using glass fiber posts than when stainless steel posts are used. Higher fracture loads were measured experimentally for teeth restored using glass fiber posts, as observed in previous works (Barjau-Escribano et al., 2006; Rodríguez-Cervantes et al., 2007). The model that was developed also predicted that the use of stainless steel and glass fiber posts to restore teeth would also result in different biomechanical performance, as the safety factor obtained for glass fiber post restorations is significantly higher than that obtained for stainless steel post restorations.

The model used to study the stress distribution inside the restored tooth proved to be suitable for studying the influence of post material on the biomechanical performance of restored teeth, since model predictions matched the experimental results obtained from the fracture strength test very well. The proposed model could be a useful tool with which to study, by means of simulations, how other post design variables influence the biomechanical performance of restored teeth.

The differences in the influence of post material on the fracture load between both post systems that were studied can be attributed to the different elastic moduli of the posts, in accordance with previous works (Barjau-Escribano et al., 2006; Rodríguez-Cervantes et al., 2007). Using a post with an elastic modulus close to that of dentin and core would prevent stress concentrations, that

weaken the restored tooth, from occurring at the post-core-cement interface.

Within the limitations of this study, we can therefore state that the use of post materials with an elastic modulus close to that of dentin and with a strength that is equal to or higher than that of dentin (as is the case of glass fiber) would be a very interesting choice for post-endodontic restoration.

ACKNOWLEDGMENTS

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Simulation of the bone filling of a dental alveolus

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ABSTRACT: After a dental extraction or loss, the healing process involves the formation of a blood clot which fills the space left by the tooth. This blood clot is progressively replaced by trabecular bone that finally fills the socket up (Li et al., 2008). In this work, a bone remodelling model based in a continuum damage-repair theory, developed by Doblaré & García (2002) was implemented to predict the bone density distribution in the alveolar cavity and its temporal evolution, in order to investigate whether this cavity is filled with bone of good quality as some clinical observations establish (Adeyemo et al., 2008) and how fast. The use of a bone remodelling scheme instead of a modelling one is justified since there is no movement of the alveolar walls, but a replacement of blood clot with bone (Kingsmill, 1999).

1 INTRODUCTION

Living materials have a very complex behavior and have been a subject of intense research. Bone tissue is a porous, heterogeneous and generally anisotropic material which adapts its microstructure and mechanical properties in accordance to the mechanical environment. The adaptive behavior of its microstructure and properties are a consequence of the process known as bone remodelling.

Many models have been proposed to predict bone mass changes in response to mechanical stimulus. Frost (1987) in his mechanostat theory postulated that bone adapts its mechanical properties according to the needed mechanical function. Bone mass, bone geometry and hence bone strength is adapted according to the every-day usage. After him, Turner (1999) proposed a new theory to address some of the flaws of the mechanostat theory. He based his theory on the assumption that bone cells react strongly to transients in their environment but eventually “accommodate” to steady state signals. He established a relaxation function to represent the bone mass change in response to external stimuli. Some of these concepts may apply to alveolar bone after a dental loss or extraction.

A certain time after tooth extraction or loss, the entire socket is filled with new bone formed by osteoblasts (Yugoshi et al., 2002). The origin of these osteoblasts is not very clear, but some authors proposed the fibroblast present in the periodontal ligament (PDL) as the progenitors (Lin et al., 1994) of such osteoblasts. These authors concluded after their studies in rats that PDL fibroblasts actively proliferate and migrate into the coagulum after tooth extraction, forming a dense connective tissue

and differentiating into osteoblasts which form new bone during socket healing.

Several stages could be distinguished in the healing process (Andreasen et al., 2007). In the first one, a coagulum is formed, once hemostasis has been established. A second stage involves the formation of granulation tissue along the socket walls and is characterized by the migration of cells like fibroblasts to the coagulum and their differentiation into osteoblasts. This granulation tissue replaces the coagulum in this stage. In the third stage the granulation tissue is replaced by the connective tissue, which in the final stage is remodeled to obtain a trabecular bone tissue (Li et al., 2008; Kingsmill, 1999; Robert & Chase, 1981).

In this work, only the final bone remodelling stage has been simulated, assuming that the connective tissue has been already formed and fills up the whole socket, with a uniform density distribution.

The remodelling model used here is based upon the model developed by Doblaré & García (2002) but now including the “Principle of Cellular Accommodation” as Turner (1999) proposes to account for the accommodation to steady state signals mentioned above. This model is employed to simulate the temporal evolution of bone density in the socket. The bone density distribution is obtained and qualitatively compared with a CT obtained from a human mandible whose first right molar have been lost and the socket filled up with bone.

Upon validation of the results this method provides a tool for predicting the density of bone in the socket a certain time after a dental loss or extraction. This is very useful in prospective implantation studies, as having a good quality of bone at the moment of implantation is fundamental for the implant short-term stability.

2 MATERIALS AND METHODS

2.1 FE model: Geometry and material properties

The FE model used here has been taken from a previous work (Martínez-Reina et al., 2007). It was meshed with linear 8-noded hexahedral elements and had a total of 77,490 elements and 88,836 nodes.

PDL and teeth are assumed to have an elastic linear isotropic behavior with the elastic constants shown in Table 1. The layer of enamel that covers the tooth above the gingival is very thin and does not affect much the global stiffness, thus it has not been considered here.

The adaptative mechanical properties of bone were simulated with a “user material” subroutine (UMAT) of ABAQUS®, based on the formulation developed by Doblaré & García (2002). The set of model's parameters used here were obtained by Martínez-Reina et al. (2007) after a sensitivity analysis performed in order to better characterize the behavior of the mandible. This formulation is summarized in section 2.3.

2.2 FE model: Boundary and loading conditions

Mastication has been simplified to be considered a pseudostatic process. Following Carter et al. (1987) the peak stresses during a cycle are responsible of bone remodelling. These peak stresses can be estimated by solving a static problem at the instant of centric occlusion, which almost coincides with the instant of maximal mastication force (Graf, 1975; Hylander, 1992). During centric occlusion the mouth is closed and the articular surfaces of condyles are pushing against the temporal articular eminence of the temporomandibular joint (TMJ).

Forces applied by masticatory muscles and boundary conditions were taken from Martínez-Reina et al. (2007). Muscle forces were applied as external loads, distributed over the insertion area of each muscle. Mastication forces are the result of the pressure distribution in the tooth-food contact. This contact has not been simulated and the problem has been simplified, as Koriotoh et al. (1992) make, by constraining the vertical displacements of the nodes of the occlusal face of the tooth that comes into contact with the food.

Thus, the reaction forces at these nodes simulate the masticatory forces.

The same procedure has been followed to simplify the contact in the TMJ: the nodes of the articular surface of the condyles are not allowed to move in the direction normal to the temporal eminence plane.

In this work, an alternating unilateral mastication pattern was considered, since this is the most usual case as reported by Manns & Diaz (1988). This pattern consists in a succession of unilateral mastications with the right molars (the load step named RM) and the left molars (LM).

A total of 500 masticatory cycles per day have been assumed and loads have been grouped, so that 500 cycles of only one type of load (RM or LM) occurs each day. Loads alternate every day, resulting in a 1 RM to 1 LM ratio. This assumption is justified by the fact that bone remodelling is a process whose typical response time is about days, allowing such grouping, first employed by Jacobs et al. (1997).

2.3 Bone remodelling model

Bone remodelling is a coupled process of old bone resorption and replacement with a newly formed tissue carried out respectively by osteoclasts and osteoblasts.

Bone resorption and apposition occur on the bone matrix surfaces. The balance of both effects gives the surface remodelling rate, $\dot{r}$, which quantifies the bone volume increment (or decrement) per unit surface and unit time. $\dot{r}$ is related to the mechanical stimulus, so that a high mechanical stimulus promotes bone formation while disuse causes bone resorption. Typical behaviors of different bones are shown in Figure 1. These curves are simplified with piecewise relations like the ones shown in Figure 2 proposed by Carpenter & Carter (2008). $\dot{r}$ is calculated by means of the mechanical stimulus Ψ_i (defined below) and the remodelling curve. This curve exhibits a zone, named “dead-zone” around a reference stimulus Ψ_i^* . If the mechanical stimulus lies within this zone neither resorption nor formation takes place.

The local mechanical stimulus is defined for every bone site using the local stress state, following Carter et al. (1987):

$$\Psi_i = \left(\sum_{i=1}^N n_i \bar{\sigma}_{ii}^m \right)^{1/m} \text{ MPa/day} \quad (1)$$

where n_i is the number of daily cycles of load i , m is an empirical exponent, N is the number of different load cases and $\bar{\sigma}_{ii}$ is the effective stress at the tissue level due to load i . Beaupré et al. (1990)

Table 1. Values of the Young's modulus, E , and the Poisson's ratio, ν , in non-remodelling materials.

	E (MPa)	ν
Dentin*	17,600	0.25
Periodontal ligamen†	3	0.45

*Craig and Peyton, 1958. †Ralph, 1982.

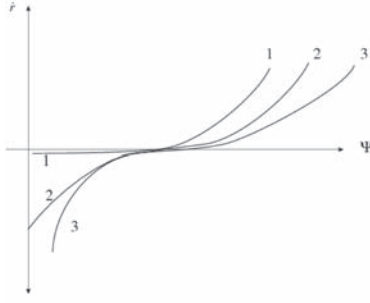


Figure 1. Remodelling behavior of different bones: 1: cranium, 2: femur periosteal, 3: femur endosteal. Taken from Beaupr  et al. (1990).

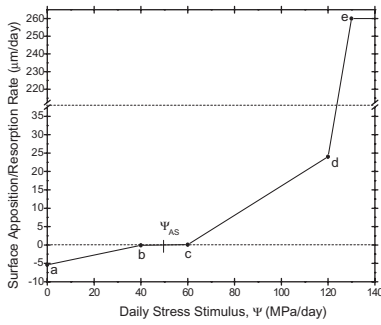


Figure 2. Simplified remodelling curve used by Carpenter & Carter (2008).

established a relationship between the apparent (macroscopic) stress, $\bar{\sigma}_i$, and $\bar{\sigma}_{ii}$ through the density of bone matrix, $\hat{\rho}$, and the apparent density, ρ :

$$\bar{\sigma}_{ii} = \left(\frac{\hat{\rho}}{\rho} \right)^2 \bar{\sigma}_i \quad (2)$$

$$\bar{\sigma}_i = \sqrt{2E(\rho)U_i(\sigma_i)}$$

where U_i is the local strain energy density of load i and E is the Young's modulus of bone, which is correlated with its apparent density following Hernandez et al. (2001):

$$E(\rho) = 3204\rho^{2.58} \quad (3)$$

The remodelling model developed by Doblar  & Garc a (2002) is based on the theory of Continuum Damage Mechanics and uses the apparent density, ρ , and the fabric tensor $\hat{\mathbf{H}}$ (directly associated to the internal microstructure and thus anisotropy) as internal variables. Bone formation and resorption criteria are defined through a pair of functions of the mechanical stimulus, g^f and g^r in the following way:

$$\begin{aligned} g^f(\mathbf{J}, \Psi_t^*, w) &\leq 0, \quad g^r(\mathbf{J}, \Psi_t^*, w) > 0, \quad \text{resorption,} \\ g^r(\mathbf{J}, \Psi_t^*, w) &\leq 0, \quad g^f(\mathbf{J}, \Psi_t^*, w) > 0, \quad \text{formation,} \\ g^f(\mathbf{J}, \Psi_t^*, w) &\leq 0, \quad g^r(\mathbf{J}, \Psi_t^*, w) \leq 0, \quad \text{dead zone,} \end{aligned} \quad (4)$$

where $\mathbf{J}$ is a tensor which depends on Ψ_t and w is the dead zone width.

The remodelling curve depends on the type of bone and its function within the skeleton. So, flat bones, with a protective mission, like the cranium has a low reference mechanical stimulus and a very small slope in resorption, much lower than load-bearing bones, like the femur (see Figure 2). Turner's Principle of Cellular Accommodation (Turner, 1999) could explain such behavior. He proposed a relaxation function to describe the changes in bone apparent density as a response to the mechanical stimulus (strain in this case):

$$\frac{d\rho}{dt} = k[S - F(S, t)] \quad (5)$$

where S is the strain and $F(S, t)$ is the reference strain (analogous to Ψ_t^* in this model) which depends on time and the applied strain, due to the process of cellular accommodation. Under steady state conditions ($S = S_0$) as $t \rightarrow \infty$ $F(S, t) \rightarrow S = S_0$, and no further change of apparent density will occur. Thus the Principle of Cellular Accommodation establishes that bones not supporting much load (like the cranium) will not be completely resorbed, as the remodelling curve would predict if Ψ_t^* were a universal value. Instead of that, Ψ_t^* is a dynamic value, which can also vary from one bone to another. Flat and protective bones will have a low value of Ψ_t^* , due to their non load-supporting history, preventing them from being resorbed.

In this work, we propose that the local reference stimulus Ψ_t^* tends to the local mechanical stimulus Ψ_t with the relaxation law:

$$\frac{d\Psi_t^*}{dt} = \varphi(\Psi_t - \Psi_t^*) \quad (6)$$

where φ represents the stimulus accommodation rate and has been assumed to be equal to 0.01 days⁻¹.

With the implementation of the Principle of Cellular Accommodation the remodelling curve (Fig. 2) is not static anymore and needs to be adjusted daily. The remodelling curve used in this study is shown in Figure 3 for a generic value of Ψ_t^* . Two points of this curve have been fixed: those ones corresponding to the maximum resorption and formation rates ("a" and "e" respectively). Points "d" and "e" of Figure 2 have been merged

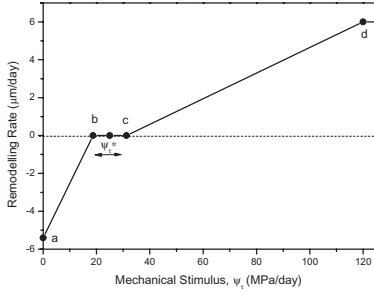


Figure 3. Remodelling curve. “a” and “e” are the fixed points of maximum remodelling rate, Ψ_t^* is variable.

into one, eliminating the second linear zone in formation. The reason is that point “e” was taken from studies of distraction osteogenesis (Lobo et al., 2004) which obviously, cannot be considered a normal situation for remodelling purposes, as it involves completely different biological mechanisms. Thus, point “d” remains at $\psi_t = 120$ MPa/day (Carpenter & Carter, 2008) and represents now the maximum bone apposition rate (saturation of the formation process). These authors proposed for point “d” a rate of bone apposition of $24 \mu\text{m/day}$, which is still too high since it corresponds to normal growth in rats (Sontag, 1992). The maximum formation rate has been estimated here as:

$$\dot{r}_{\max} = V_{BMU} \cdot f_{bio} \quad (7)$$

where f_{bio} is the maximum BMUs activation frequency (García-Aznar et al., 2005) and V_{BMU} is the volume of tissue formed by a BMU. In cortical bone, the volume was approximated by Hernandez (2001) with a hollow cylinder:

$$V_{BMU} = \frac{\pi}{4} (d_0^2 - d_H^2) \cdot v_{BMU} \cdot \sigma_L \quad (8)$$

where d_0 and d_H are the diameters of the osteon (external diameter) and the haversian canal (internal diameter) respectively, v_{BMU} is the BMU progression rate and σ_L is its life span (García-Aznar et al., 2005). In trabecular bone the volume formed by a BMU was estimated as half of a cylinder with an elliptic section (Parfitt, 1994):

$$V_{BMU} = \frac{\pi}{4} \cdot d_e \cdot d_{BMU} \cdot v_{BMU} \cdot \sigma_L \quad (9)$$

where d_e and d_{BMU} are the erosion depth and the width of the resorption cavity respectively. This procedure applies also to estimate the resorption rate, since Equation 7 gives the maximum tissue formed or resorbed by a BMU. The remodelling

process involves both, apposition and resorption, and a balance must be made to work out the net effect. So Equation 7 is valid just in the case that only one of these processes takes place. Thus, Equation 7 provides an upper bound of the maximum, either in formation or in resorption.

The values taken for the parameters in Equations 7–9 are shown in Table 2. With those values the maximum remodelling rate estimated for cortical and trabecular bone are $6.0 \mu\text{m/day}$ and $1.2 \mu\text{m/day}$ respectively. They are quite different, but, in both cases, much lower than the value adopted by Carpenter & Carter (2008). The value $6.0 \mu\text{m/day}$ has been adopted here for point “d” and saturation.

The resorption rate for total disuse ($\psi_t = 0$ MPa/day) was set at $-5.4 \mu\text{m/day}$, which was the maximum resorption rate that occurred in the immobilized forelimbs of old beagles (Jaworski et al., 1980, Cowin et al., 1985). Interestingly, this value is very close to the rate assessed before for cortical bone.

Points “b” and “c” define the dead zone, whose semi-width is fixed at 25% of Ψ_t^* (Beaupré et al., 1990). Finally, the slopes of the linear sections must be worked out for every reference stimulus. Figure 3 shows the remodelling curve finally adopted.

2.4 Other features of the simulations

The initial density and mechanical properties of the rest of the mandible (not the socket) were taken from Martínez-Reina et al. (2007) (see Fig. 4).

The extraction of the first right molar produces the formation of a blood clot, replaced during the initial stages of the healing process by a connective tissue, as it was explained before. The starting point of the simulations performed in this work is precisely the end of the healing stage after which the connective tissue is formed. To simplify the behavior of the socket, the connective tissue has not been considered. Instead of that, the socket has been assumed to be filled with bone from the beginning. An initial uniform density of 0.05 g/cm^3

Table 2. Parameters to estimated V_{BMU} for a cortical bone.

Parameters	Nominal value
d_0^a	$200 \mu\text{m}$
d_H^a	$40 \mu\text{m}$
d_e^a	$49.1 \mu\text{m}$
d_{BMU}^a	$152 \mu\text{m}$
v_{BMU}^a	$40 \mu\text{m/day}$
σ_L^a	100 day
f_{bio}^b	0.05

^aParfitt, 1994. ^bGarcía-Aznar et al., 2005.

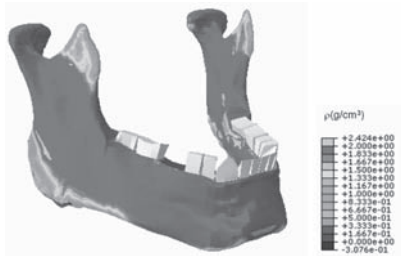


Figure 4. Initial apparent density distribution obtained by Martínez-Reina et al. (2007). The socket is filled with bone of density 0.05 g/cm^3 .

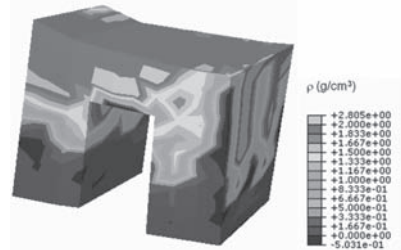


Figure 5. Density distribution of the alveolar socket after 400 days of activity.

has been assigned to the bone in the socket, so that it has similar mechanical properties to those of the connective tissue. Bone of density 0.05 g/cm^3 would have 1.5 MPa of Young's modulus in comparison to 2 MPa of connective tissue as was obtained by Hori & Lewis, 1982.

In accordance with the Principle of Cellular Accommodation, the reference stimulus assigned to the connective tissue filling the socket (or the bone that substitutes it, in the simplification made here) is initially zero, since this tissue has just been formed and has not supported any mechanical stimulus yet. In the rest of the mandible it has been assumed that a steady state has been reached prior to dental loss and thus, the initial reference stimulus distribution equals the mechanical stimulus distribution obtained by Martínez-Reina et al. (2007).

3 RESULTS

A total of 400 days of mastication have been simulated. This results to be enough time to get a remodelling equilibrium in the bone that fills the socket. In fact, the apparent density of this bone changes very little after the first 200 days.

The apparent density of the whole mandible after 400 days is not shown since there are no appreciable changes. It must be noted that the starting point was a remodelling equilibrium situation, except in the socket. Consequently, only local changes around the zone of the extracted tooth have occurred.

Figure 5 shows the apparent density distribution in the bone that fills the socket after 400 days and Figure 6 compares the density distribution obtained numerically (left) with a CT (right) taken at the site of extraction.

4 DISCUSSION

As stated above there is no density changes in the socket after 200 days of activity which is in good

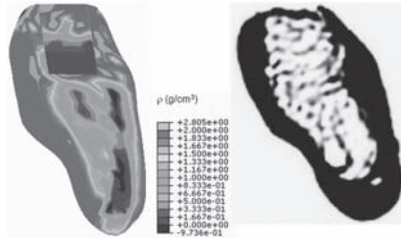


Figure 6. Left: Density distribution in the section of the mandible at the site of extraction after 400 days of activity. Right: CT computer tomography taken at the same section.

agreement with the normal usage of waiting 3 months after extraction for the implantation surgery.

Comparing the numerical results of the density distribution with the CT, a reasonable qualitative match can be noticed: the socket is filled with a bone of a high density at the top and low density at the bottom, although these kind of "lid" is too thick in the results. Moreover, our model predicts the existence of a zone of high density below the socket which is not real. Resorption must occur there but our model does not predict it.

There seems to be another inconsistency of our results at the bottom of the section (Fig. 6), where a dense and thick layer of cortical bone is suggested by the CT. In fact, the density in this zone (estimated by Hounsfield units) is the lowest of the cortical layer, despite the appearance, thus, results are not far from reality.

5 CONCLUSIONS

In this work we have proposed a method for estimating the temporal evolution of apparent density in the bone that fills the alveolar cavity after a dental loss or extraction. The interest of this study is to provide a tool to investigate whether that bone has enough quality for supporting initial stability to a dental implant. A previously developed bone

remodelling model has been adapted to simulate the particular behavior of this tissue which is initially connective but is remodeled to end up as a trabecular bone of low density with a “lid” of a more dense bone, closing the former hollow.

Results are very promising, as it could be checked out by comparing the density distribution obtained numerically with a CT taken from the mandible modeled here, whose first right molar was also missing. An interesting result is that the remodelling equilibrium is achieved in the socket after 200 days approximately, which is in agreement with some clinical observations.

However, there are still some points that need to be studied more carefully: resorption below the socket has not been achieved; a sensitivity analysis studying the influence of the accommodation rate, ϕ , must be done and the most important one, further and more detailed clinical validation is called for.

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Histological evaluation of pulp tissue and periodontal regeneration in autogenous tooth transplantation

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ABSTRACT: **Objective:** The aim of this study was to evaluate the pulp tissues and periodontal regeneration, in autogenous tooth transplantation, with application of Emdogain® in the socket.

Methods: The study group comprised 3 Beagle dogs, in which 24 incisors and premolars were transplanted to the recipient sockets after mechanical preparation. The Emdogain® group (E), the saline group (S) and the control group (C) contained 12, 12 and 6 teeth, respectively. Clinical examinations were performed every week and were sacrificed 9 weeks later. Subsequently, decalcified sections were prepared for routine histological immunohistochemical evaluation. Ordinal scores were analyzed using the t-student test ($p = 0.05$).

Results: All the transplanted teeth survived. No statistically significant difference was found among the complete healing in all treatment groups ($p = 0.303$). The mean occurrence of inflammatory ($p = 0.015$) and replacement root resorption was high with saline solution, compared with Emdogain® group, with significant difference ($p = 0.015$ and $p = 0.041$, respectively).

Conclusions: The Emdogain® does not seem to enhance the regeneration of periodontal and pulp tissues, but decrease the inflammatory and the replacement root resorption.

1 INTRODUCTION

Autogenous tooth transplantation could be an alternative way to restore a missing tooth if there is a suitable donor tooth available.^{1,2} Successful tooth transplantation depends upon the optimal and uneventful healing of periodontium.^{3,4}

It depends upon the vitality of remaining periodontal ligament cells in the donor root, the shape and the site of the recipient socket and the vascularity of the recipient bed.⁵⁻⁸

To improve nutrition and preserve cell activity in these tissues Nethander et al⁶ and Katayama et al⁷ suggested that teeth should be transplanted to the sockets with regeneration tissues, which reduce the root resorption following transplantation.

Enamel matrix proteins from Hertwig's epithelial root sheath, have been shown to regenerate acellular extrinsic cementum in monkey's. Emdogain® (EMD), a commercial preparation of porcine enamel matrix derivative, has shown a promise in periodontal regeneration in periodontal defects in humans.⁹⁻¹¹ EMD should stimulate some tissues promoting the development and regeneration of acellular cementum.^{11,12} In vitro studies also implicated EMD in the induction of human PDL

fibroblasts and stimulate the increased production of alkaline phosphatase and transforming growth factors β_1 (TGF- β_1) in PDL cells.¹³ Thus hypothetically, in transplanted teeth wounds, EMD might be able to effect periodontal ligament regeneration, inhibiting root resorption. The purpose of this study was histologically to evaluate the effect of Emdogain® gel on periodontal healing in transplanted teeth of dogs.

2 MATERIAL AND METHODS

2.1 Animal preparation

Before start of the experiment, application was made to the *Direcção de Serviços de Meios de Defesa da Saúde, Bem-estar e Alimentação Animal da Direcção Geral de Veterinária (Portugal)*. The animal experimental procedures were performed in *Estação Zootécnica Nacional (EZN)—Instituto de Tecnologia Biomédica- Santarém- Portugal*, in accordance with the International Guiding Principles for Animal Research. The animals are aquired in *Universidade de Córdoba—Espanha*, and maintained in the EZN during the entire experimental period were daily observed by the

medical staff of the EZN. Non carious and periodontal sound mature incisors and premolars from 3 male Beagle dog (age 5 months; body weight 11.73 ± 1.13 kg) were selected. All experimental procedures were performed with the animals under general anesthesia, accomplished by pre-anesthetic sedation with 0.05 mg/kg of body weight of acepromazine (Calmivet®- Vetoquinol- Lure- France) administered intramuscular and anesthetic induction with 10 mg/kg of body weight administered intravenous of thiopental (Pentotal® -Queluz de Baixo- Portugal). Finally, intubation was made and maintenance of anesthesia was a mixture of oxygen and 1–2% of isoflurane (Isoflo, Veterinaria Esteve, Barcelona- Spain). Throughout the duration of general anesthesia, the dog's received normal saline solution intravenously.

2.2 Experimental procedures

All selected teeth were extracted as atraumatically as possible, under aseptic conditions. The alveolar sockets were amplified with a bur Ø 3.5 mm to receive the incisors and Ø 4.2 mm to receive the premolars (Straumann- Basel- 4052 Switzerland), under continuous irrigation with normal saline solution.

2.3 Treatment groups

The teeth were extracted as atraumatically as possible. The roots and sockets were gently rinsed with 5 ml each of normal saline solution immediately before the following treatment protocols.

Group E – The teeth were transplanted after 0.1 ml of Emdogain® gel was dispensed from a blunt needle syringe onto the rot surface.

Group S – The teeth were transplanted after 0.1 ml normal saline solution was dispensed from a blunt needle syringe onto the rot surface.

After the transplantation the teeth were splinted with vycril 3/0.^{14,15} The animals were given standard soft diet with hot water. Immediately after transplantation and for 5 subsequent days, the animals were given enrofloxacin (5 mg/kg of body weight) and 0.01 mg/kg of body weight of buprenorphine once daily.

2.3.1 Specimen processing

The animals were killed 9 weeks after transplantation. The dog's were deeply anesthetized with an overdose of intravenous pentobarbital at 100 mg/kg of body weight. After the dissection to expose the carotid vein, perfusion was performed with 40 ml of 4% paraformaldehyde in phosphate buffer (pH 7.4). Jaw blocks containing the transplanted teeth were resected and fixed in the same fixative, decalcified in 50% formic acid and 20% sodium citrate and

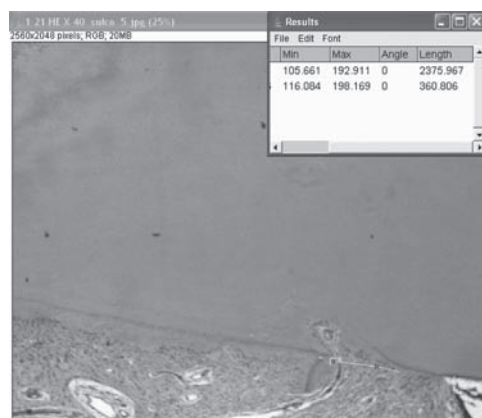


Figure 1. Delineation of the root resorption for the histomorphometric analysis. (HE × 40).

embedded in paraffin. The paraffin embedded blocks were subsequently sectioned longitudinally in buccolingual direction at a thickness of 5 µm, corresponding at the root canal. The section that was technically the best was stained with hematoxylin and eosin or Van Giesson.¹⁶ The other sections were stained immunohistochemically for keratin MNF116 or Vimentine.¹⁷

2.3.2 Histomorphometric evaluation

Histomorphometric analysis was performed with the aid of the software ImageJ 1.30 (Image Processing and Analysis in Java—National Institute of Mental Health, Bethesda, Maryland, USA) and at 40x magnification. The histology of the periodontium was classified by using the classification as follows: percentage of complete healing; percentage of superficial root resorption; percentage of inflammatory root resorption; percentage of replacement root resorption (Fig. 1).

2.3.3 Statistical analysis

The percentage of each histologic classification for each root and each treatment group was calculated. The differences among the groups were statistically compared by using the Mann-Witney test. The significance level was set at 5%.

3 RESULTS

The dogs tolerated the operative procedures well and their behavior did not change. No teeth were lost and the number of transplanted teeth was 24. The results of the histomorphometric analysis for each treatment group are presented in Table 1. In teeth transplanted with Emdogain®, the mean occurrence of complete healing was high (93%),

compared with saline solution (81%), but there was no significant difference ($p = 0.303$). The mean occurrence of replacement resorption was high with saline solution (9%), compared with Emdogain® (2%), with significant difference ($p = 0.041$). The mean occurrence of superficial root resorption was high with saline solution (8%), compared with Emdogain (3%), but there was no significant difference ($p = 0.818$). The mean occurrence of inflammatory root resorption was high with saline solution (11%), compared with Emdogain (0%), with significant difference ($p = 0.015$). In the histological observations at 9 weeks after transplantation, the space of the periodontal ligament was reestablished (Fig. 2).

We can see by MNF116, in the periodontal ligament the epithelial cells of Malassez were observed (Fig. 3).

In the pulp we can see a hard tissue like bone that was well characterized with Vimentine (Fig. 4).

Group E- transplanted teeth with Emdogain®; Group S- transplanted teeth with saline solution.

Table 1. Periodontal healing pattern for different groups and statistical comparison using Mann-Witney test ($P = 0.05$).

	E (n = 12)	S (n = 12)
Complete healing ($P = 0.303$)	93%	81%
Mean (%)		
Inflammatory root resorption ($P = 0.015$)	0%	11%
Mean (%)		
Replacement root resorption ($P = 0.041$)	2%	9%
Mean (%)		
Superficial root resorption ($P = 0.0818$)	3%	8%
Mean (%)		

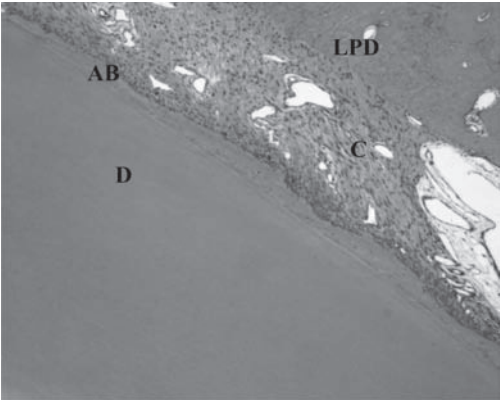


Figure 2. Histological photograph of the experimental root surface with complete healing (Dog1, teeth 21-Group S). Alveolar Bone (AB); Periodontal Ligament (LPD); Cement (C); Dentin (D) (HE $\times 200$).

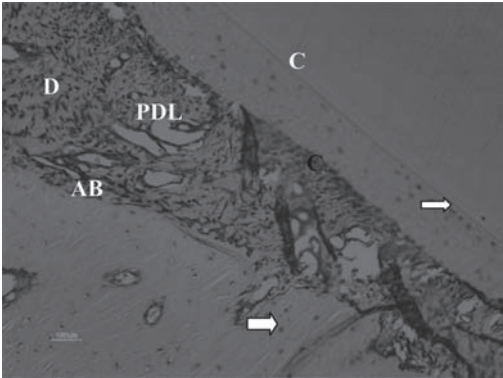


Figure 3. Histological photograph of the experimental teeth (Group E). Periodontal Ligament (PDL); Cement (C); Dentin (D). Osteocits (big arrow) and cementocits (small arrow), with great expression of Vimentin, (VM $\times 100$).

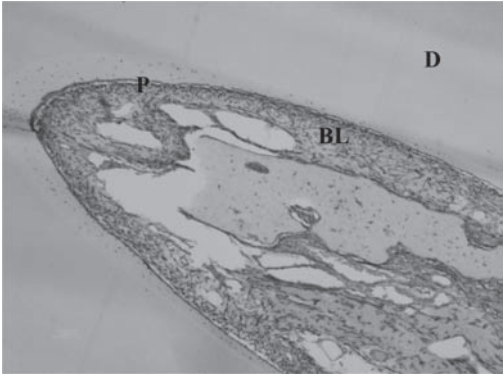


Figure 4. Histological photograph of the experimental teeth (Group E). Pulp (P) with bone-like matrix (BL), with great expression of Vimentin, Dentine (D). (VM $\times 40$).

4 DISCUSSION

Studies on the complications of root resorption after transplantation teeth have been performed based on model in monkeys, rats and dogs.^{5,6,18,19} In this study, we used transplant teeth in dogs, after 0.1 ml of Emdogain gel or 0.1 ml normal saline solution dispensed from a blunt needle syringe onto the root surface. The periodontal ligament has demonstrated a remarkable capacity for repair and regeneration. All transplanted teeth demonstrated a similar progression of regeneration. In addition the osteoblast-like cells lining the newly formed alveolar bone, suggest that osteogenic cells in periodontal ligament might proliferate and differentiate into osteoblast-like cells. Emdogain was found to be able stimulation of alkaline phosphatase

activity and expression of bone matrix proteins in osteoblast without contact direct with cells.^{13,20} It has been also demonstrated the capacity of healing after injury of pulp as a pulp-capping material.²¹ The mean occurrence of replacement resorption was high with saline solution (9%), compared with Emdogain® (2%), with significant difference ($p = 0.041$). The mean occurrence of superficial root resorption was high with saline solution (8%), compared with Emdogain® (3%), but there was no significant difference ($p = 0.818$).

TGF- β 1 inhibits the production of inflammatory cytokines and promotes the production of extracellular matrix and proliferation of osteoblasts and odontoblasts.²² These effects result in the suppression of inflammatory reactions and promote wound healing. This effect may be responsible for the complete periodontal healing compared with saline solution. In all experimental groups Emdogain®, the mean occurrence of complete healing was high (93%), compared with saline solution (81%), but there was no significant difference ($p = 0.303$). The mean occurrence of inflammatory root resorption was high with saline solution (11%), compared with Emdogain® (0%), with significant difference ($p = 0.015$). The histological observations at 9 weeks after transplantation demonstrated that the space of the periodontal ligament was reestablished and the epithelial cells of Malassez were observed. The hard tissue like bone, well characterized with Vimentine was found in the pulp tissue representing the capacity to form bone-like tissues.

5 CONCLUSION

In conclusion, this animal study demonstrated that there were differences in wound healing process between Emdogain® and saline solution in transplanted teeth. Our results also indicate that periodontal ligament tissue contains osteoprogenitor cells that have the ability to participate in alveolar bone regeneration. These results have clinical implications and suggest that if non-functional teeth as third molars are available, autotransplantation may be considered as treatment of choice. Autotransplantation can be an alternative to dental implants in some patients in whom dental implants become impossible due to inadequate bone support and patients in growing stages.^{1-4,19,23}

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Methods for assessing dental wear in bruxism

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ABSTRACT: The dental wear is a problem that bothers patients and clinical dentists because of the structure loss is irreversible and the damage can trigger pain. The task of to monitor the dental wear is essential to prevent possible progressions and to plan an intervention in this process. The research present in this paper can help to analysis of methodologies that can contribute to the monitoring and management of bruxism and the resulting tooth loss. This study analyzed both profilometry and tested the replicability of new materials for construction of the samples was also developed a protocol for the wear and last for me has developed a program of Computer Vision based on photographic images to serve an estimated wear. The results are motivating and suggest new avenues of study and monitoring of bruxism.

1 INTRODUCTION

1.1 *Bruxism*

Bruxism is the habit of clenching or grinding tooth (Lavigne, 2005). Most of the population (85–90%) will, at some point of life, have this habit to a certain degree (Bader & Lavigne, 2000). It causes irreversible lost of teeth structure, pain and damage to oralfacial structures.

1.2 *Profilometry*

Profilometry is a very common method and also sophisticated, as it makes it possible to extract data

from superficial topography in nanometric scale. In this method, a very delicate stylus slides over a surface and its vertical dislocations are converted on a two-dimensional graphic representing its profile. The assembly of several images of profiles results in a tridimensional image. Data are recorded as peak and valleys distribution (Figure 1).

In this study, the followed parameters were chosen to represent the surface characteristics:

1. S_a ($\mu\text{m}/\mu\text{m}$): average roughness;
2. S_q ($\mu\text{m}/\mu\text{m}$): quadratic average roughness,
3. S_{dq} ($\mu\text{m}/\mu\text{m}$): quadratic average slope,
4. S_{sc} ($1/\mu\text{m}$): quadratic average curvature,

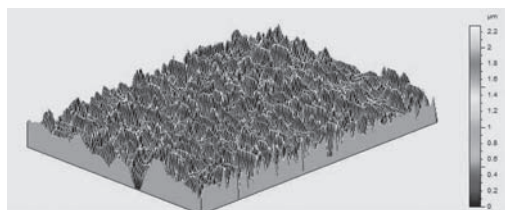


Figure 1. Rugosity profile of a tooth.

5. Ssk: asymmetry coefficient for the distribution height curve relative to the average,
6. Sku: curtosis coefficient.

The first and second coefficients provide an insight on the height's dispersion in relation to the medium surface of the surface, while the next two help in characterizing their shapes. Asymmetry and curtosis coefficients measure the departure of the obtained surface from a standard normal distribution (Mummery, 1992).

1.3 Protocol

The proposed protocol was based on Davies (2002), undergoing some changes based on Lavigne (2005). It also includes a questionnaire with questions for systematization of a clinic evaluation. The type of bruxism was determined by a set of parameters: facets analysis, pain symptoms (presence and localization), presence of noise, volunteer's report or others report. In the chosen population, 100% do not make use of splints, devices used in the treatment of temporomandibular joint disorders.

1.4 Computer vision

In this part of the study, the purpose is the automatic measurement of the levels of dental wear using patient pictures taken over the time. The method uses pictures taken from several mouth views of volunteer or pictures of models taken at different times. The output is an estimate of dental wear occurred over those times. The development and implementation was in partnership with the Computation Science Department of Federal University of Minas Gerais.

2 METHODOLOGY

2.1 Replica

In (Bastos, 2004), the author suggested a direct *in vivo* application of profilometry with the use of replicas after a validation stage. In the present study, 8 extracted teeth were used, after donation from volunteers under the procedures approved by the

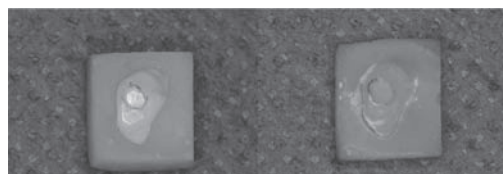


Figure 2. Samples: first one the original teeth and the second in low viscosity resin.

Bioethics Committee. To improve the measuring procedure, teeth were sectioned along their long axis and then placed parallel to the surface containing the area of interest (wear facet). The tooth fragment was then stabilized in acrylic resin. Later it was molded with addition-curing silicone and the obtained models filled with two materials that are not usually used in Dentistry for similar applications: epoxy resin and low viscosity dental resin. The choice of these materials was due to the fact that they have the required fluidity to copy details of the geometry and provide specimen with sufficient hardness to support the profilometry procedure (Figure 2).

Profilometry tests (Hommel Tester T4000 from Hommelwerke GmbH) were performed over selected surfaces of the eight teeth and their replicas. Comparison between the original tooth and its replica was made using 2-tailed Student t test for paired samples, with an alpha of 0,05, described by Fisher. (Sampaio, 2007). The comparison was done using average values of the parameters, both for tooth/epoxy replica and tooth/resin replica.

2.2 Protocol

The evaluations of 17 volunteers (nine male and eight female) were accomplished in 18 months with 6 months intervals between each evaluation. The age range was 20–33 years and there were a total of 3 evaluations for each individual. The exclusion criterion was presence of mobility and systemic or local serious changes.

2.3 Computer vision

Computer Vision is the area of Science dedicated to obtaining models of the world based on digital images. We define our problem as a need of monitoring tooth wear in a large time interval using just images of the mouth of a patient. This is a complicated problem, and can not be solved by simple techniques. First, we need a sophisticated camera with a high resolution CCD (or CMOS) sensor, because the wear we intend to measure may be very small (less than millimeters). Another problem is the illumination in the instant of the image capture, which constitutes an important feature in some computer

vision problems. As the tooth wear occurs in large intervals of time (in general, months or even years), it becomes impossible to keep the illumination conditions invariable in a common clinical dentist.

In order to facilitate the image processing in the computer vision algorithms, it is desirable also to keep the camera (and the patient) static between the acquisitions of these images, but this can also be a problem, once the time interval between images is large. Even if the dentist trying to frame the image in order to match with images taken months later, the final result will be a variation of the original one, due to camera parameters such as focal distance and lens distortion.

In the present methodology, we try to estimate the tooth wear of a patient comparing images taken in different instant times and with different illumination and camera view conditions. With this, it is possible to estimate the tooth wear from images taken by a non-calibrated camera in a non-calibrated environment, which represents a common clinical dentist procedure.

The proposed method is divided in three steps: (I) a re-projection of the images using projective geometry, (II) segmentation of each tooth and (III) the estimation of the tooth wear using the profile.

2.3.1 Problem geometry

In order to be able to compare directly images taken from different points of view, we must first align the image plane on the space. This will permit to correct the effects of scale between the images (due to the distance or zoom variation of the camera), and the rotation in relation to the scene. This alignment can be made by estimating the function that represents the geometric variation between these two images in the space. This function is generally called projective transformation.

To avoid the need of complex methods in the moment of image acquisition from patients or models, a method was purposed in this study that aim to equal geometrically the taken images in different instants.

2.3.2 Projective transformation

The projective transformation is widely used in computer vision as it is able to describe different views of the same scene in the 3D space (Gracias, 2000). This transformation is modulated by the homographic function (2D projective transformation), defined as a function H (Forsyth & Ponce, 2002). It needs correlated points in the same plane in 3D space in different images.

2.3.3 SIFT (Scale Invariant Feature Transform)

The Scale Invariant Feature Transform (SIFT) is an efficient filter to extract and describe key points of images (Lowe, 2004). It is a robust method to

treatment of noise, bad illumination, occlusion and minor changes in viewpoint, diminishing mismatches in points matching. SIFT is not perfect, it allow outliers, i.e., error in the point matching. The RANSAC are used to robust estimation. It is an iterative method to estimate parameters of a mathematical model from a set of observed data which contains outliers. Finally, after the outliers removal using RANSAC is estimated the homography is applied to correct the point of view in images. After this step, the tooth profile is determined using segmentation in the images.

2.3.4 Wear identification

Once were two images with the same geometric representations, I and $\hat{I}$, the next step is the profile identification of the grinded teeth, which will give the dental loss estimative related with time.

2.3.5 Image segmentation

Segmentation refers to the process to divide a digital image in regions or objects. The result of image segmentation is a set of regions or a set of contours extracted from the image. As a result, each of the pixels in the same region is similar in any characteristic, such as color, intensity, texture. Adjacent regions must have significant differences with respect its same characteristic.

The used procedure was based on the manual threshold I which one to each pixel in the image is classified as Teeth or Non teeth (according to the intensity values). There are methods that allows automatic choice of this threshold but in this case, only for experimental testing, these values were founded manually.

2.3.6 Teeth profile calculation

With the binary images, the next step is to extract the teeth profile. The profile represents the teeth contour in the image in function of their lines and columns. Once established the profile for each image, the next step is to estimate the dental wear throughout the considered time.

2.3.7 Estimation of tooth wear

The estimative is the difference in pixels area and lately, after conversion of pixels value to dimensions from the real world. To minimize the differences in those profiles, an optimization method known as *Levenberg–Marquardt* Method was used.

3 RESULTS AND DISCUSSION

3.1 Replica

In Table 1 it is possible to verify that copies in resin as well as in epoxy do not present difference statistically from the original tooth at 95% of

Table 1. Value of t 7GL, 2- taile alpha 5% = 2,365.

Parameters	t2 epoxy	t2 Flow resin
Sa	1,334814954	-0,360652229
Sq	0,944408645	1,5988499100
Sdq	-1,100191092	-0,2177840310
Ssc	-1,619292963	-1,4637218480
Ssk	-1,448656316	-2,0211579390
Sku	0,2128843230	0,4556827240

significance. That means that is possible to use models in epoxy and in Flow resin models to study superficial texture. In this case, the *in vivo* study is possible and superficial changes that are not sensed by other methods in a short period of the time could be detected.

An advantage would be the detection of dental wear in initial condition without macro loss, another advantage would be studies of the parameters and their relationship to wear mechanisms. In this paper, only the steps of testing the applicability and materials evaluation were made, in the future, a study with patient's replicas is planned to try to predict the development of dental wear and also an association to the other methodologies indicated on this paper.

3.2 Protocol

In Table 2, it is possible to verify the facets distribution and severity related to dentin exposition. The major proportion of teeth affected by facets were canines. The results indicate a raise in number of facets as well as in number of tooth with dentin exposure.

As to the severity and progression, a significant difference was found at the end of the study with 82% of the volunteers presenting a progressive and symptomatic wear situation.

3.3 Image tests

3.3.1 With direct images

To test the program were used pictures taken over the time. The tooth in question was a canine. To determine the transformation matrix between those two images related points among two images were calculated using SIFT, as previously described. Relations can be visualized in Fig. 3, blue lines representing the correspondent points linking the two images.

Applying the transformation from the homographic matrix it was possible to create a new representation of the second image (with wear) that represents a perspective more similar to the first one, as the two images were taken from the same angle.

The next step is the identification of the teeth profile. In this case, first a manual selection of a

Table 2. Facets distribution and dentin exposition.

Measurement	Number of		% Dentin exposition	
	Restorations	Facets	Yes	No
First	83	83	71,1	28,9
Second	87	91	74,7	25,3
Third	90	96	79,5	20,6

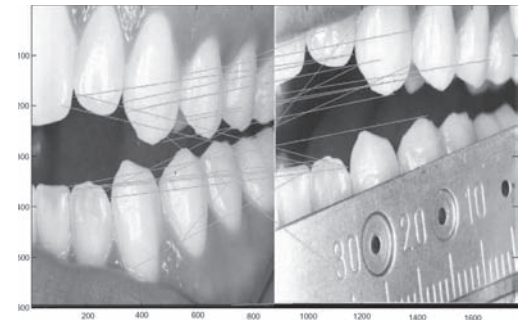


Figure 3. Use of SIFT in test with volunteer.

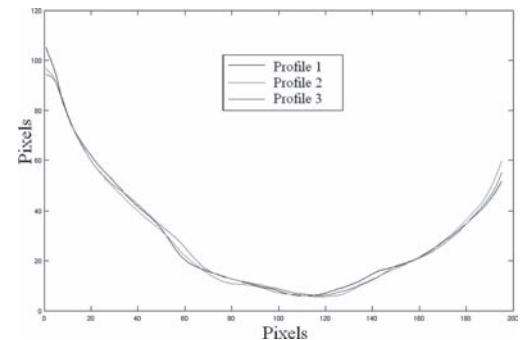


Figure 4. Final result to direct image.

section from the images of this singular tooth was done. In the final step was established a comparison between the two profile curves. After the dislocation, scale and rotation of the second profile in relation to the first one was achieved (Fig. 4). Then, the profiles of the same teeth took at different times can be compared directly in scale (number of pixels) of the image.

Calculating the wear area under the curves the value of $0,1 \text{ mm}^2$ was found when comparing the first to the second and $0,18 \text{ mm}^2$ when comparing the third image to the first. At present, it is not possible to separate noise from real wear. In the presented case, the volunteer was included in the protocol phase and his clinical evaluation

in 18 months reinforce the image tests results. He presented progression with painful symptoms, propagation and increase of cracks and increment of dentin exposition.

3.3.2 Tests with replica

At least, tests with dental replicas were made using pictures of replicas from volunteer, achieved in a similar way that was achieved as in profilometry (Figures 5 and 6). Again the selected tooth was a canine.

All the sequence of procedures of image treatment were followed and applied here. Final result can be seen in Figure 7.

The value in this case was 1,2 mm². Again, is possible to notice a significant difference, indicating teeth loss due to dental grinding.

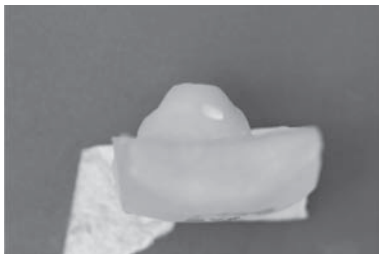


Figure 5. Initial replica.



Figure 6. Final replica after 6 months.

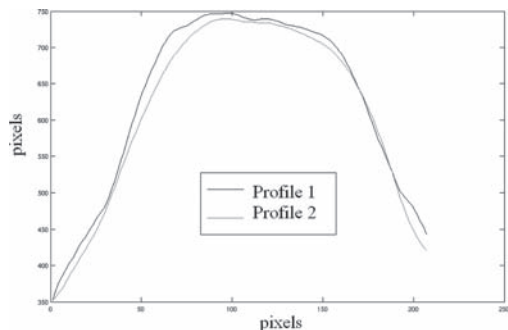


Figure 7. Final result to replica case.

The preliminary results indicate very positive motivation for further studies. There are several ways to reduce noise and improve the program. The biggest advantage of this method is the automatic character that makes it easier the caption of image in dentistry.

4 CONCLUSIONS

From these results it is possible to summarize the developments of this paper as follows:

1. The possibility of using replicas to make possible the study of human tooth via profilometry was tested and succeeded;
2. The improvement in the technique of models production makes it easier to apply profilometry;
3. The test with resin Flow shows a new application to restoration material in dentistry in surface studies;
4. Protocols which helps on the prediction and progression of Bruxism where applied on a population and indicating a progressive wear situation on the studied population;
5. It was presented and tested a Computational Vision program that can estimate dental wear in a very simple and potentially reliable way;
6. The applicability and results about these three methods were discussed.

5 FUTURE WORKS

In the future, a wide study with the 3 methodologies applied connected is desired to try to predict, evaluate and estimate dental wear. The combination of the three methods has a potential to provide a comprehensive method for following up and understand dental wear mechanisms.

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Three dimensional skeletal muscle tissue modeling

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ABSTRACT: Human skeletal muscle tissues present a very complex mechanical behavior being active, quasi-incompressible, transversely isotropic and hyper-elastic. In this paper, a three-dimensional finite element muscle model is proposed to characterize the complex behavior of muscle tissues. The muscle is regarded as the ground substance matrix embedded with muscle fibers. The substance matrix is modeled as an isotropic material. The constitutive model for the muscle fibers is based on Hill's 3-element model. In addition, volume conservation is imposed to ensure the incompressibility of the muscle tissues. The proposed muscle material model was implemented into the finite-element solver LS-DYNA® by means of user-defined material subroutines. The performance of the proposed model was evaluated by comparing the simulated results with the published experimental data from rabbit muscle studies. The work is summarized in this paper, showing that the proposed constitutive muscle model is able to model both the active and the passive muscle tissue behavior for strains below failure.

1 INTRODUCTION

Skeletal muscle tissues play an important role in human body movement. They can generate voluntary forces leading to body motion and provide strength and protection to the skeleton.

Skeletal muscle tissues have a complex mechanical behavior, with the fundamental property of being able to contract without any mechanical influence from outside. This contraction is controlled by the neural electrical stimulation. When given a neural stimulation, the actin and myosin filaments freely slide relative each other, with the calcium ion concentration increased. Thus, cross-bridges are formed between the actin and myosin filaments (Guyton & Hall 2000). The passive behavior of muscle tissues can be characterized by non-linear hyperelastic stress-strain relations (Fung 1981). In addition, the muscle tissues are essentially incompressible and can be treated as transversely isotropic material when the fibers are paralleled to each other.

One of the first mathematical muscle models was proposed by Hill (1938). However, his phenomenological model is only one dimensional (1D) and consists of three elements: a contractile element (CE) in series with an elastic element (SEE), both in parallel with a passive elastic element (PE) (Fig. 1). Thus, this model is unsuitable for prediction of the deformation of skeletal muscles with complex geometry under complex loading conditions.

Some 3-Dimensional (3D) muscle models have been developed over the past decades to overcome the limitations of the 1D Hill's muscle model. One way of developing a 3D model is to extend Hill's 1D muscle model to 3D. The approach is to combine an isotropic ground substance matrix with the active muscle fibers, which are described by Hill's 1D model, e.g. in Martins et al. (1998); Johansson et al. (2000); Oomens et al. (2003); Blemker et al. (2005); Martins et al. (2006) and Tang et al. (2009). Following the same idea, a general 3D skeletal muscle model based on the concept of fiber-reinforced composite is proposed in this paper, in which the substance ground matrix behavior is predicted by a strain energy function proposed by Humphrey & Yin (1987), whilst the active and passive behaviors of the muscle fibers are characterized by Hill's 1D equations. Several numerical tests are performed to validate and verify the proposed model.

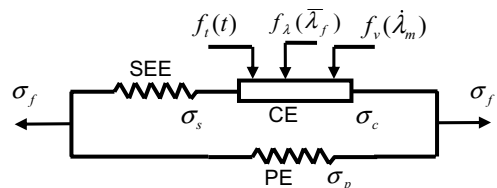


Figure 1. Muscle model (Hill's model).

2 THE MUSCLE CONSTITUTIVE RELATION

The muscle constitutive relation is derived through the strain energy approach and the formulations are within the framework of nonlinear solid mechanics (Truesdell & Noll 2004).

2.1 The strain energy density function

The muscle is regarded as a fiber-reinforced composite comprising a ground substance matrix and muscle fibers (Fig. 2). We assume the fibers are distributed in parallel and have a single direction.

The strain energy in the muscle is given by:

$$U = U_I(\bar{I}_1^C) + U_f(\bar{\lambda}_f, \lambda_s) + U_J(J) \quad (1)$$

where

$$U_I(\bar{I}_1^C) = c\{\exp[b(\bar{I}_1^C - 3)] - 1\} \quad (2)$$

is the strain energy stored in the isotropic matrix; $\bar{I}_1^C$ = first invariant of the right Cauchy-Green strain tensor with the volume change eliminated; and b & c = material parameters.

$$U_f(\bar{\lambda}_f, \lambda_s) = \int_1^{\bar{\lambda}_f} [\sigma_s(\lambda, \lambda_s) + \sigma_p(\lambda)] d\lambda \quad (3)$$

is the strain energy stored in the muscle fibres; $\bar{\lambda}_f$ = fiber stretch ratio with the volume change eliminated; λ_s = stretch ratio in SEE; λ = fiber stretch ratio; $\sigma_s(\lambda, \lambda_s)$ = stress produced in SEE; and $\sigma_p(\lambda)$ = stress produced in PE;

$$U_J(J) = \frac{1}{D}(J - 1)^2 \quad (4)$$

is the strain energy associated with the volume change; J = Jacobian of the deformation gradient; and D = compressibility constant.

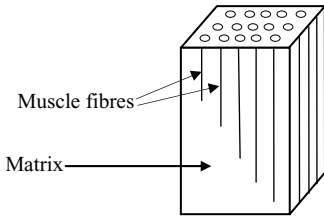


Figure 2. Diagram illustration of a parallel-fibered muscle.

2.2 Stress produced in the series element (SEE)

Based on the experiments of Pinto & Fung (1973) on the papillary muscle of a rabbit heart, a recurrence relation is proposed to express the stress produced in SEE (Fung Y.C. 1981):

$${}^{t+\Delta t}\sigma_s = e^{\alpha\Delta\lambda_s} ({}^t\sigma_s + \beta) - \beta \quad (5)$$

with

$${}^t\sigma_s = \beta[e^{\alpha({}^t\lambda_s - 1)} - 1] \quad (6)$$

where α & β = material constants.

Equation (5) contains one unknown, namely $\Delta\lambda_s$, and this can be solved using the method proposed by Kojic et al. (1998). The idea is to set up a non-linear equation with the unknown $\Delta\lambda_s$ by utilizing the stresses relationship between CE and SEE, i.e. the stress in CE is equal to the stress in SEE at any moment.

$${}^{t+\Delta t}\sigma_m = {}^{t+\Delta t}\sigma_s \quad (7)$$

2.3 Stress produced in the contractile element (CE)

The stress produced in CE is given by:

$${}^{t+\Delta t}\sigma_m = \sigma_0 \cdot f_t(t + \Delta t) \cdot f_\lambda(\bar{\lambda}_f) \cdot f_v(\dot{\lambda}_m) \quad (8)$$

where σ_0 = maximal isometric stress; $f_t(t + \Delta t)$ = muscle activation function; $f_\lambda(\bar{\lambda}_f)$ = muscle stress-stretch function; and $f_v(\dot{\lambda}_m)$ = muscle stress-velocity function.

2.3.1 The activation function

The activation behavior of the muscle is quite complex and still under research. In this paper, an exponential function, which has also been used by Meier & Blickhan (2000), is adopted:

$$f_t(t) = \begin{cases} n_1, & \text{if } t < t_0 \\ n_1 + (n_2 - n_1) \cdot h_t(t, t_0), & \text{if } t_0 < t < t_1 \\ n_1 + (n_2 - n_1) \cdot h_t(t_1, t_0) - [(n_2 - n_1) \cdot h_t(t_1, t_0)] \cdot h_t(t, t_1), & \text{if } t > t_1 \end{cases} \quad (9)$$

with

$$h_t(t_i, t_b) = \{1 - \exp[-S \cdot (t_i - t_b)]\} \quad (10)$$

where n_1 = activation level before and after the activation; n_2 = activation level during the

activation; t_0 = activation time; t_1 = deactivation time; and S = exponential factor. When modelling single muscle fibres, the magnitude of parameter S is related to the rate of the chemical processes. For modelling large muscles, S represents the time dependent recruitment of different motor units.

2.3.2 The function of stretch

The forces developed in a muscle depend on muscle's sarcomere length. In order to find the relationship between the isometric tension and the sarcomere length, Gordon et al. (Gordon et al. 1966) conducted a series of experiments on a single fibre of frog skeletal muscle and found a piecewise linear isometric tension versus length dependency. In our model, a smooth quadratic function proposed by Blemker et al. (2005) is used to approximate Gordon's experimental curve. This quadratic function has also been used by Böl & Reese (2008). It has the following form:

$$f_\lambda({}^t\bar{\lambda}_f) = \begin{cases} 0, & \text{if } {}^t\bar{\lambda}_f / \lambda_{\text{opt}} < 0.4 \\ 9({}^t\bar{\lambda}_f / \lambda_{\text{opt}} - 0.4)^2, & \text{if } 0.4 \leq {}^t\bar{\lambda}_f / \lambda_{\text{opt}} < 0.6 \\ 1 - 4(1 - {}^t\bar{\lambda}_f / \lambda_{\text{opt}})^2, & \text{if } 0.6 \leq {}^t\bar{\lambda}_f / \lambda_{\text{opt}} < 1.4 \\ 9({}^t\bar{\lambda}_f / \lambda_{\text{opt}} - 1.6)^2, & \text{if } 1.4 \leq {}^t\bar{\lambda}_f / \lambda_{\text{opt}} < 1.6 \\ 0, & \text{if } {}^t\bar{\lambda}_f / \lambda_{\text{opt}} \geq 1.6 \end{cases} \quad (11)$$

where λ_{opt} = optimal fibre stretch.

Compared to the piecewise linear function used by Tang et al. (2009), this smooth function has the big advantage that it reduces the parameter inputs from five to one.

2.3.3 The function of velocity

It is well-known that the force generated in the muscle during contraction is highly dependent on its velocity of contraction (Hill 1970). In 1938, Hill proposed a hyperbolic relation between the muscle force and the velocity, which is still used by scientists today.

However, Hill's force-velocity function is restricted to the concentric and isometric contraction of the muscle. The tension-velocity relation for muscle lengthening was first characterized in the form of an equation by Otten (1987). Later, this hyperbolic equation was used by Van Leeuwen (1991).

The stress-velocity function incorporated into our model is derived from these functions and given by:

$$f_v(\dot{\lambda}_m) = \begin{cases} \frac{1 - \dot{\lambda}_m / \dot{\lambda}_m^{\min}}{1 + k_c \dot{\lambda}_m / \dot{\lambda}_m^{\min}}, & \text{if } \dot{\lambda}_m \leq 0 \\ d - (d - 1) \frac{1 + \dot{\lambda}_m / \dot{\lambda}_m^{\min}}{1 - k_c k_e \dot{\lambda}_m / \dot{\lambda}_m^{\min}}, & \text{if } \dot{\lambda}_m > 0 \end{cases} \quad (12)$$

where k_c & k_e = shape parameters of the hyperbolic curves, which control the curvature of the curve; d = offset of the eccentric function; $\dot{\lambda}_m^{\min}$ = stretch rate in the CE; and $\dot{\lambda}_m^{\min}$ = minimum stretch rate.

By using equations (5), (7) & (8), the following non-linear equation for solving the stress increment in SEE, $\Delta\lambda_s$, can be obtained:

$$f(\Delta\lambda_s) = (\alpha_2 + \alpha_3 \Delta\lambda_s) e^{\alpha_4 \Delta\lambda_s} - \alpha_4 \Delta\lambda_s - \alpha_5 = 0 \quad (13)$$

where, in case of muscle shortening

$$\alpha_2 = ({}^t\sigma_s + \beta) \left(1 + \frac{k_c \cdot \alpha_1}{\dot{\lambda}_m^{\min} \cdot \Delta t} \right) \quad (14)$$

$$\alpha_3 = -({}^t\sigma_s + \beta) \frac{k \cdot k_c}{\dot{\lambda}_m^{\min} \cdot \Delta t} \quad (15)$$

$$\alpha_4 = -\frac{\beta \cdot k_c - {}^t f_\lambda(\bar{\lambda}_f) \cdot f_t(t + \Delta t)}{\dot{\lambda}_m^{\min} \cdot \Delta t} k \quad (16)$$

$$\alpha_5 = \beta + {}^t f_\lambda(\bar{\lambda}_f) \cdot f_t(t + \Delta t) - \frac{{}^t f_\lambda(\bar{\lambda}_f) \cdot f_t(t + \Delta t) - \beta \cdot k_c}{\dot{\lambda}_m^{\min} \cdot \Delta t} \alpha_1 \quad (17)$$

and in case of muscle lengthening

$$\alpha_2 = ({}^t\sigma_s + \beta) \left(1 - \frac{k_e \cdot k_c \cdot \alpha_1}{\dot{\lambda}_m^{\min} \cdot \Delta t} \right) \quad (18)$$

$$\alpha_3 = ({}^t\sigma_s + \beta) \frac{k \cdot k_e \cdot k_c}{\dot{\lambda}_m^{\min} \cdot \Delta t} \quad (19)$$

$$\alpha_4 = \frac{\beta \cdot k_e \cdot k_c + {}^t f_\lambda(\bar{\lambda}_f) \cdot f_t(t + \Delta t) \times (d \cdot k_e \cdot k_c + d - 1)}{\dot{\lambda}_m^{\min} \cdot \Delta t} k \quad (20)$$

$$\alpha_5 = \beta + {}^t f_\lambda(\bar{\lambda}_f) \cdot f_t(t + \Delta t) - \frac{{}^t f_\lambda(\bar{\lambda}_f) \cdot f_t(t + \Delta t) \cdot (1 - d - d \cdot k_e \cdot k_c) - \beta \cdot k_e \cdot k_c}{\dot{\lambda}_m^{\min} \cdot \Delta t} \alpha_1 \quad (21)$$

with

$$\alpha_1 = (1 + k) e^{t + \Delta t} \bar{\lambda}_f - {}^t \lambda_m - k {}^t \lambda_s \quad (22)$$

Once $\Delta\lambda_s$ is solved, the stress in SEE can be obtained by using equation (5).

2.4 Stress produced in the parallel element (PE)

When the muscle is not activated, the forces in CE and SEE are zero. The force in PE is positive when the muscle is stretched and null when the muscle is compressed. Based on the experimental test (Chen & Zelter 1992), the stress in PE can be expressed as:

$${}^{t+\Delta t}\sigma_p = \sigma_0 f_{PE}({}^{t+\Delta t}\bar{\lambda}_f) \quad (23)$$

with

$$f_{PE}({}^{t+\Delta t}\bar{\lambda}_f) = \begin{cases} A \cdot ({}^{t+\Delta t}\bar{\lambda}_f - 1)^2, & \text{if } {}^{t+\Delta t}\bar{\lambda}_f > 1 \\ 0, & \text{otherwise} \end{cases} \quad (24)$$

where A = material parameter.

With the stress expressions from equations (5) & (23), the strain energy produced in the muscle fibers can now be obtained from equation (3). This way the total strain energy in the composite is solved.

Table 1. Material parameters.

Description	Parameter	Value	References	
Stress in matrix	b	23.46	Humphrey & Yin (1987)	
	c (N/m ²)	379.0		
Stress in SEE	α	10	Pinto & Fung (1973)	
	β (N/m ²)	1.0e5		
Stress in PE	A	4.0	Chen & Zelter (1992) Zajac F.E. (1989)	
	σ_0 (N/m ²)	7.0e5		
Stress in CE	$f_i(t)$	S (s ⁻¹)	50	Meier & Blickhan (2000)
	$f_v(\dot{\lambda}_m)$	k_c	5	
		k_c	5	Close (1964), Böhl & Reese (2008)
		d	1.5	
		$\lambda_c^{\min}$	-17	Myers et al. (1995)
	$f_\lambda(\bar{\lambda}_f)$	λ_{opt}	1.05	
		k	0.3	Fung (1981)
Compressibility constant	D (m ² /N)	1.0e-9	Martins et al. (2006)	

The Cauchy stress tensor can be derived from the strain energy (Belytschko 2000).

There are 14 material parameters in our model, the values of which are based on the available data found in the literature. These values and the corresponding references are listed in Table 1.

3 MODEL VALIDATION AND VERIFICATION

The proposed muscle constitutive model was implemented into LS-DYNA® by means of user-defined material subroutines (UMAT). Several validation tests have been performed. For the passive and activated elongation tests, the simulation results are compared to the experimental data from the New Zealand white rabbit hind leg muscle tibialis anterior (Myers et al. 1995, 1998, Davis et al. 2003). An isometric contraction simulation was also performed, even though no test data was available, for the purpose of model verification.

A simple muscle FE model shown in Figure 3 was used for the validation tests. The length of the muscle is 5.0e-2 meters. The radius in the smallest cross section is 9.0e-3 meter and in the largest cross section it is 1.75e-2 meter. The initial direction of the parallel distributed fiber was chosen to be along z direction. The material parameters listed in Table 1 were adopted.

3.1 Passive elongation

In the passive elongation test, the muscle was pulled quasi-statically from its rest length, while the muscle was not activated. The engineering stress and strain relationship was obtained from the simulation results and plotted in Figure 4, with the available experimental results included for comparison. Figure 4 shows reasonably good agreement between the experimental data and our passive elongation simulation results.

3.2 Activated elongation

The activated elongation simulation is divided into two stages. In the first stage, the muscle was held

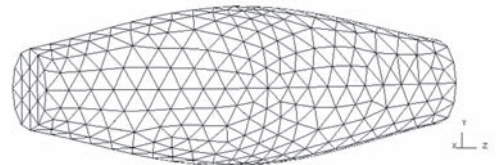


Figure 3. Muscle FE mesh.

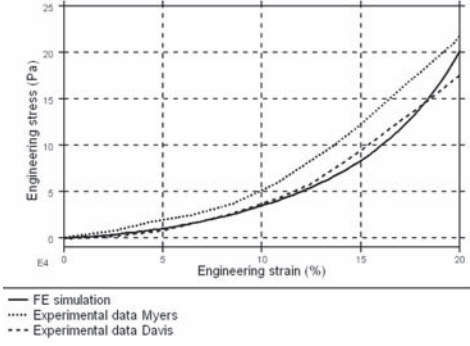


Figure 4. Engineering stress versus strain curves from passive elongation simulation.

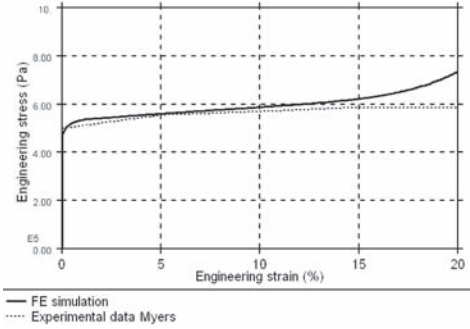


Figure 5. Engineering stress versus strain curves from activated elongation simulation.

constant in length while being stimulated for 0.5 second, at the end of which the muscle had reached full activation. In the second stage, the muscle was pulled quasi-statically while the full activation was maintained. The stress response predicted by our model is in accordance with the experimental data up to 15% engineering strain (Fig. 5).

3.3 Isometric contraction

In this example, the muscle was subjected to an isometric contraction, thus its two ends were clamped during the simulation. A neural excitation with the amplitude of 1.0 was applied at 0.1 second and kept constant for 0.3 second, after which the neural excitation was gradually reduced to zero. Thus, in the activation function definition, we give $n_1 = 0.0$; $n_2 = 1.0$; $t_0 = 0.1$; $t_1 = 0.4$. The activation curve is shown in Figure 6.

The Cauchy stress components in the z axis direction have the following relation:

$$\sigma_{33} = \sigma_{CE33} + \sigma_{VOL33} + \sigma_{I33} + \sigma_{PE33} \quad (25)$$

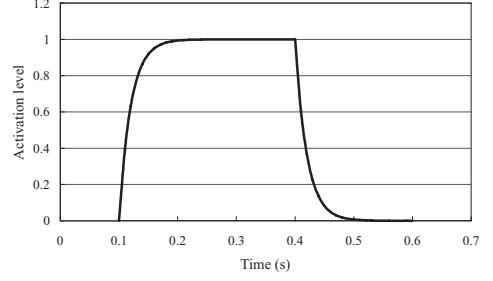


Figure 6. Activation function for the isometric contraction simulation.

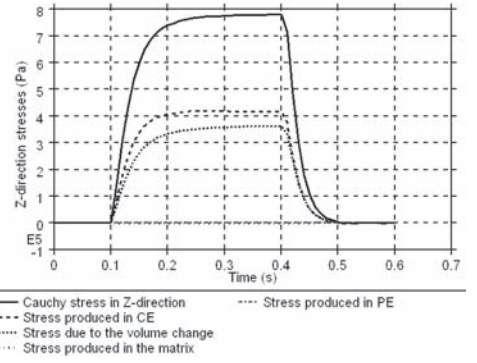


Figure 7. Z-direction stresses versus time at a node in the centre of the muscle for the isometric contraction simulation.

where σ_{33} = Cauchy stress component in z direction; σ_{CE33} = Cauchy stress component produced in CE in z direction; σ_{VOL33} = Cauchy stress component due to the volume change in z direction; σ_{I33} = Cauchy stress component produced in the matrix in z direction; and σ_{PE33} = Cauchy stress component produced in PE in z direction.

The five stress components versus time at a node located in the middle part of the muscle are plotted in Figure 7. Since the length of the muscle is held constant during the simulation, the stresses produced in the parallel element and the isotropic substance matrix are zero, as expected. The time variation of the other three stresses reflect the evolution of the activation function, i.e. they have the same shapes as the activation function. These theoretical predictions are verified by the numerical results (Fig. 7).

4 CONCLUSIONS

In this paper, a 3D Hill-based skeletal muscle model has been developed to characterize its complex

mechanical behavior. The proposed constitutive model has been implemented into the non-linear FE program, LS-DYNA® by means of user defined subroutines. The model was validated and verified by demonstrating that it is capable to simulate the passive and active muscle behaviors during both the shortening and lengthening movements.

We are currently using the described constitutive muscle model on a FE model of a human face, for the purpose of facial movement simulations, e.g. facial expressions mimic, as most of the facial movements are induced by muscle stimulations. This work is currently under development.

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New formulations for space provision and bone regeneration

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ABSTRACT: Bone grafting materials are frequently used in several clinical indications related to endodontic, periodontal or implant surgery to improve tissue regeneration. Additionally, there is a trend in developing biologic modalities that may enhance bone healing of specific sites. With this regard, a synthetic cell-binding peptide (P-15) incorporated in a scaffold (Anorganic Bovine Matrix [ABM]) has been frequently used to facilitate the attachment, migration, and differentiation of osteoblastic cells. Recently, a new presentation of this type of material has been developed with the purpose of improving its handling, controlling particle migration and optimizing clinical efficacy. It has been suggested that creating a more homogenous interparticle spacing, which is crucial for a proper cellular and vascular colonization, could promote a faster bone regeneration with relevant clinical benefits. The aim of this study was to compare the performance of different bone grafts constitute by ABM/P-15 particles with and without different carriers. A demineralized rabbit allograft in granular form, suspended in a carrier was additionally investigated. **Materials and Methods:** 32 adult, male, rabbits were divided into four test groups and one positive control group (ABM/P-15 particulate without carrier). Critical size defects in the distal femur were filled with ABM/P-15 particles (Group I) and with ABM/P-15 particles using distinct carriers: carboxymethylcellulose and glycerol (Group II); sodium hyaluronate (Hy) hydrogel (Group III), and lyophilized plugs (Group IV). Additionally, demineralized bone allograft suspended in sodium Hy was used in Group V. Block sections were harvested at 2 and 4 weeks for histological processing. **Results and discussion:** Group II presented a lowest percentage of new bone formation, in both evaluation periods. On the contrary, at 2 weeks, Group IV presented more new bone formation than group I and III, nevertheless these differences were not statistically significant ($p > 0.05$). On the other hand, Group V presented a significant increase in new bone formation relatively to group I, without any carrier ($p < 0.05$). At 4 weeks, Group IV presented a significant increase in new bone formation when compared with group I ($p < 0.05$) and there were no significant differences between groups V, III and I ($p < 0.05$). Between 2 and 4 weeks, there were only statistical differences in intragroup comparison for groups I and IV. **Conclusions:** These results show the influence of different formulations of bone grafts in providing an adequate scaffold for the support of the regenerative process and emphasize the importance of the type of carrier in the three-dimensional distribution of particles and space provision in new bone formation, especially in early periods. Among the carriers tested, the sodium hyaluronate (either with the ABM/P-15 or with the DBM particles) seems to be the one that improved the most bone healing/regeneration process as well as the rate of graft resorption. Furthermore, the lyophilized form exhibited the best performance.

1 INTRODUCTION

The regeneration of lost or resorbed bone tissue is one of the major concerns and challenges of the oral health professionals, and requires most often the application of a biomaterial bone graft.

To stimulate bone regeneration, several biomimetic strategies have been designed using bone

matrix analogous materials enriched with biological modulators (Guerra 2003; McAllister 2007). According to this perspective, a particulate material has been developed, composed of an anorganic bovine-derived matrix (ABM) linked to a synthetic peptide (P-15), ⁷⁶⁶GTPGPQGIAGQRGVV⁷⁸⁰. This peptide, constituted by a sequence of 15 amino acids of Type I collagen, represents a binding

cellular domain, specifically involved in the migration, adhesion, and proliferation of osteoblasts and fibroblasts (Bhatnagar et al. 1999).

Recently, in an attempt to improve clinical handling of this particulate bone graft material and to control the migration of particles, new formulations have been developed through the association of carriers (Bhatnagar 2004; Nguyen et al. 2003). These carriers are used to preserve a three-dimensional homogeneous distribution of the graft material in the osseous defect, maintaining a convenient interparticle spacing (Guerra 2005).

Moreover, a direct injectable application, without the need of hydration and compactation, would represent a qualitative step in the clinical manipulation of this bone graft (Matos et al. 2007; Nguyen et al. 2003). Nevertheless, the developed formulations have a limited level of scientific evidence regarding regenerative potentialities.

The present experimental study aims to evaluating the process of bone healing, particularly bone regeneration and osseointegration, of the ABM/P-15 graft particles in a contained bone defect. For that, different particulate formulations without carrier and with different types of carriers and particle concentrations were tested. Three carriers were used: an hydrogel of carboxymethylcellulose and glycerol (CMC), a sodium hyaluronate hydrogel (Hy) and a lyophilized sodium hyaluronate hydrogel (Hy lio). Additionally, an allograft consisting of demineralised rabbit cortical bone with particles suspended in a sodium hyaluronate carrier (DBM/Hy) was investigated.

2 MATERIALS AND METHODS

2.1 Animals

This pre-clinical study used the rabbit as the animal model following the experimental protocol approved by the Portuguese Authority—DGV (*Direcção Geral de Veterinária*). The animals were housed and manipulated according to the National Legislation (1005/92 de 23 de October; n° 1131/97 de 7 de November).

32 rabbits, New Zealand White, male adults with a medium weight of 3.8 ± 0.4 Kg, were used. The results were evaluated in two healing periods, corresponding to 2 and 4 weeks postoperative.

2.2 Surgical procedures

General anaesthesia was reached by intramuscular injection of ketamine and xylazine at a dose of 35 mg/Kg and 2 mg/Kg, respectively, and maintained during the intervention through an intravenous administration of ketamine (50 mg/ml) as needed. Both hind limbs were shaved, prepped with

povidone-iodine solution and draped. An incision of approximately 1.5 centimetres was made over the distal femur in the lateral aspect of the knee joint. Blunt dissection through the underlying muscles exposed the periosteum which was elevated to identify the epiphysis. Care was taken to avoid creating communication with the knee joint cavity. A cylindrical defect was drilled with irrigation using increasing size drills to a final 5 mm diameter by 10 mm depth through the lateral cortex of the distal femur and extending to, but not through, the opposite medial cortex. The defect was placed within the epiphysis, which mainly contains cancellous bone, and avoiding the epiphyseal plate and bone marrow cavity of the diaphysis. Following meticulous rinse with sterile saline for bone debris, one of the materials was placed into the defect site (typical volume of graft material placed in defect was approximately 0,2 cc) or left untreated.

Two defects were created per rabbit in the femur of opposite limbs, establishing the so-called critical size defect (Chan et al. 2002; Oonishi et al. 2000).

Each defect received randomly one of the materials or remained empty as a negative control. Only 2 negative controls per group were used, since previous studies (Guerra 2005; Matos 2008) using this model have demonstrated limited bone regeneration in empty defects.

The Critical size defects in the distal femur were filled with:

- ABM/P-15 (Group I)—anorganic bovine-derived mineral (1.2 gm/cc) with fifteen amino acid in particulate form with a particle size range of 250–420 μm , and a surface area of 1.0 m^2/g .
The peptide, ⁷⁶⁶GTPGPQGIAGQRGVV⁷⁸⁰ (P-15), is synthesized by solid phase procedures and adsorbed on ABM in a saturable manner (Bhatnagar et al. 1999).
- ABM/P-15/CMC (Group II)—anorganic bovine derived mineral (0.82 gm/cc) with fifteen amino acid suspended in an inert hydrogel (carboxymethylcellulose and glycerol) formulation (51% particulate content).
- ABM/P-15/Hy (Group III)—anorganic bovine derived mineral (0.51 gm ABM/P-15/cc) with fifteen amino acid suspended in a solution of sodium hyaluronate hydrogel (Hy) (37.5% particulate content).
- ABM/P-15/Hylio (Group IV)—anorganic bovine derived mineral (0.82 gm ABM/P-15/cc) with fifteen amino acid suspended in an inert hydrogel (sodium hyaluronate) formulation lyophilized in the form of 4 mm $\times$ 5 mm plugs (85.6% particulate content).
- DBM/Hy (Group V)—demineralized bone allograft suspended in sodium hyaluronate (93% allograft content).

All this materials were supplied by CeraPedics L.L.C. (Lakewood, CO, USA).

The wound was meticulously closed in layers with resorbable sutures for the periosteum and non-resorbable for the skin sutures. A critical size defect model was established through bilateral femoral condylar defects (5 mm diameter by 10 mm depth). This type of experimental model represents a suitable way of evaluating the contribution of graft materials to bone regeneration.

Butorphanol tartrate (0,2 mg/Kg body weight, s.c, 2 days) was administered for immediate post-surgery pain control and a long acting amoxycillin was given as a single dose antibiotic (1 mg/Kg, s.c).

2.3 Healing periods

The 32 animals were divided into two subgroups of 16 animals each with the following healing periods: 2 weeks and 4 weeks. At the end of each healing periods, the animals were sacrificed by induction of deep anesthesia followed by intravenous sodium pentobarbital euthanasia. Bone harvest was done en bloc with a power saw and soft tissue carefully dissected away from the defect site for histological preparation.

2.4 Histological preparations

The histological tissue processing was prepared according to a non-decalcified technique, using the Exakt® system, which allows high quality slide sections with residual morphological distortions and artefacts. This protocol was carried out at the Hard Tissue Laboratory of the Dentistry Department of the Faculty of Medicine of the Coimbra University.

The histological evaluation concerned qualitative morphologic and quantitative morphometric analysis using light microscopy slides. In the histomorphometric analysis the following parameters were evaluated: a) percentage of new bone formation (% nb); b) percentage of graft particles (% part); c) percentage of new bone related to the graft particles (nb/part %); d) and defect resolution (% (nb + part)/total area of defect *i.e.* new bone plus graft). Statistical analysis used the Kruskal-Wallis test for intergroup comparison and the Mann-Whitney test for intragroup comparison at a 95% level of significance ($P \leq 0.05$).

3 RESULTS AND DISCUSSION

3.1 Clinical results

All rabbits recovered with no response to the surgical procedure or to the graft materials. During healing, none of the animals showed signs of postoperative complications.

3.2 Histologic and histomorphometric evaluation

Histological analysis is considered the gold standard for evaluation of the new bone growth in bony defects.

Qualitative histologic observations and histomorphometric results are respectively presented in Figure 1 and Table 1 that is graphically depicted in Graph 1. Histological sections of the defects receiving no graft material (negative control sites) showed minimal bone growth. These defects, presented residual amount of new bone ($7,1 \pm 2,4\%$ e $4,14 \pm 2,2\%$, for 2 and 4 weeks) with a repairing pattern of fibrous healing, which was consistent with the critical size of the defect.

All the implanted material formulations showed good biocompatibility due to the absence of adverse tissue reactions and anatomo-morphological alterations of adjacent tissues.

Bone defects filled with the graft materials evidenced much greater amount of new bone formation compared with unfilled defects (negative control).

Regarding the above results, the following comments can be made. The defects filled with the ABM/P-15 particles (Group I) without any carrier showed an early and considerable new bone formation ($20,2 \pm 3,0\%$ and $27,6 \pm 3,1\%$ for 2 and 4 weeks), from the periphery to the centre of the defect lining and interconnecting graft particles. This osteoconductive matrix coupled with P-15 promoted the synthesis and deposition of a mineralized matrix, initially characterized as a woven bone that after remodelling substitution generated a more organized and lamellar tissue. Nevertheless, the persistence of several areas of immature bone tissue (Fig. 1A) surrounding some graft particles can still be observe after a period of 4 weeks.

The ABM/P-15/CMC (Group II) presented an anomalous particle migration, which determined an unpredictable osteoconduction, compromising the formation (Matos 2008) and maturation of a robust mineralized matrix (Fig. 1B). In this group there was a significant reduction in new bone formation ($12,3 \pm 2,4\%$ and $13,2 \pm 4,5\%$ for 2 and 4 weeks) compared with other groups, in both evaluation periods.

The ABM/P-15/Hy (Group III) showed a low particle concentration also associated with a tendency for particle migration. New bone ($19,0 \pm 3,7\%$ and $23,0 \pm 4,1\%$ for 2 and 4 weeks) deposition followed particle pattern distribution. Indeed bone trabeculae of lamellar structure were generally associated with higher particle concentration areas in the periphery of the defect.

On the other hand, lyophilized ABM/P-15/Hy (Group IV) showed the best biological performance, regarding quality and quantity of new bone

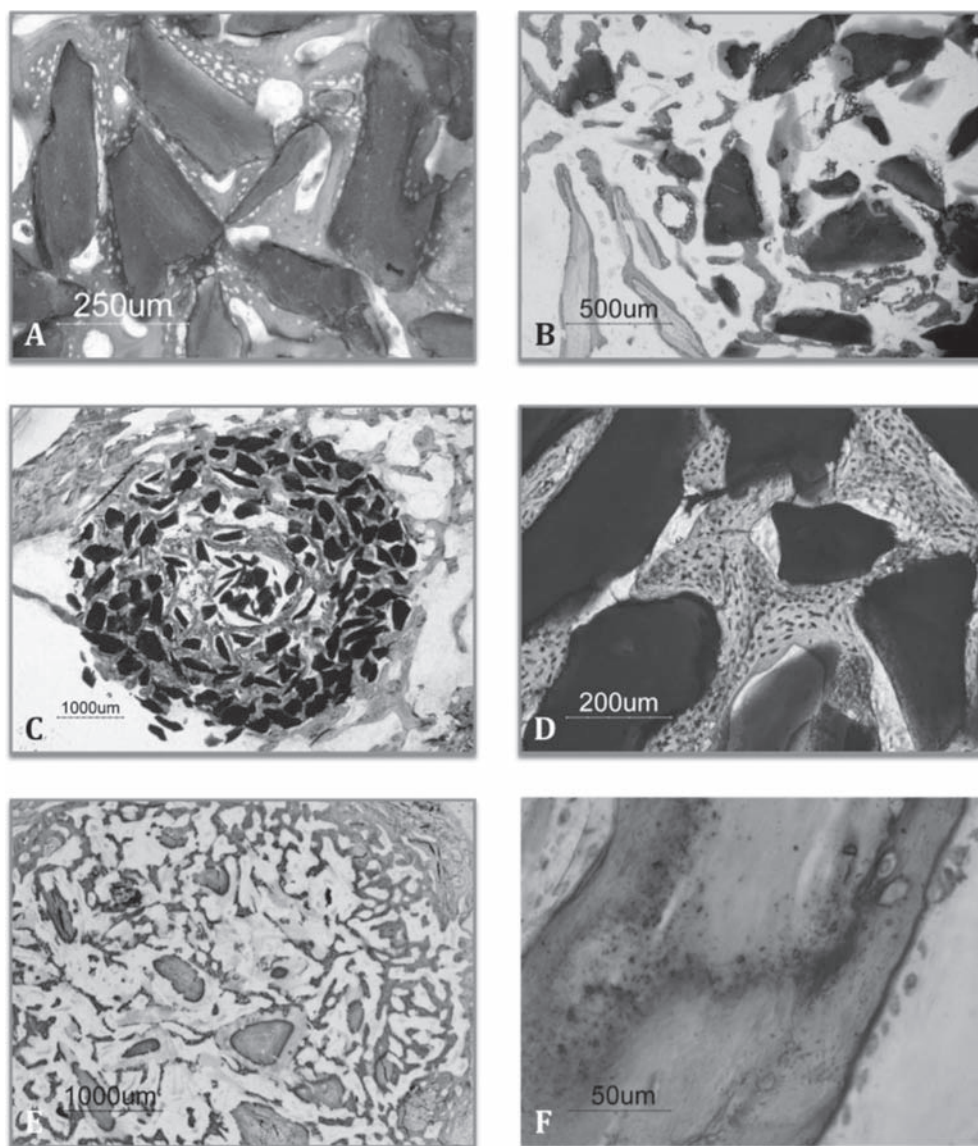


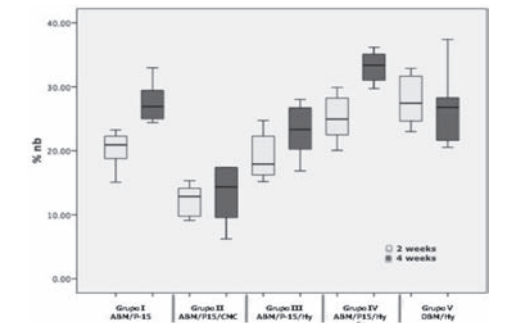
Figure 1. Microphotographs of undecalcified ground sections of bone defects. A) Group I (4 weeks)—Numerous and osteointegrated particles can be observed. However, the presence of several areas of woven bone on graft particles should be highlighted. B) Group II (4 weeks)—The peripheral area of a bone defect exhibits a low concentration of particles and bone trabeculae. C) Group IV (4 weeks)—The presence of extensive bone trabeculae network, lining and connecting the graft particles, progressing towards the central area is evident. D) Group IV (4 weeks)—Remarkable amount of thick lamellar trabeculae, coating many particles can be seen. E) Group V (2 weeks)—The formation of an extensive woven bone trabeculae network is notorious. F) Group V (4 weeks)—This image shows of a recalcification process on a DBM particle. This recalcification, is characterized by the presence of mineral deposition spots in a formerly demineralized bone matrix. Note that most of the surface of the graft particle is covered by new bone.

formation ($25.0 \pm 3\%$ e $33.1 \pm 2.4\%$, for 2 and 4 weeks), resulting in a remarkable amount of thick lamellar trabeculae (Fig. 1C and D). Concerning this particular formulation, can it concluded

that, despite being the same particle and carrier composition of the Group III, the lyophilized form originated significant differences, allowing a more suitable particles distribution and

Table 1. Histomorphometric analysis: Means $\pm$ SD of volume percent at 2 and 4 weeks for new bone (% nb), particles (Part%), new bone related to the graft particles (% nb/par) and defect resolution per healing period (% (nb+par)/total area of defect).

2 Weeks	% part	% nb	% nb/part	% (nb+par)/Total
Group I(ABM/P-15)	43.7 $\pm$ 6.1	20.2 $\pm$ 3.0	47.5 $\pm$ 12.5	63.9 $\pm$ 4.9
Group II (ABM/P-15/CMC)	17.3 $\pm$ 6.5	12.3 $\pm$ 2.4	77.1 $\pm$ 26.8	29.7 $\pm$ 7.7
Group III (ABM/P-15 Hy)	14.3 $\pm$ 3.9	19.0 $\pm$ 3.7	134.9 $\pm$ 22.7	33.1 $\pm$ 7.1
Group IV (ABM/P-15/Hy lio)	26.6 $\pm$ 3.2	25.0 $\pm$ 3.7	97.0 $\pm$ 14.5	51.1 $\pm$ 5.4
Group V (DBM/Hy)	26.1 $\pm$ 5.1	27.8 $\pm$ 3.9	112.2 $\pm$ 34	53.9 $\pm$ 2.7
4 Weeks	% part	% nb	% nb/part	% (nb+par)/Total
Group I(ABM/P-15)	43.8 $\pm$ 5.9	27.6 $\pm$ 3.1	64.0 $\pm$ 11.6	69.4 $\pm$ 4.5
Group II (ABM/P-15/CMC)	14.5 $\pm$ 3.1	13.2 $\pm$ 4.5	90.5 $\pm$ 26.6	27.7 $\pm$ 7.0
Group III (ABM/P-15 Hy)	18.2 $\pm$ 4.3	23.0 $\pm$ 4.1	132.9 $\pm$ 40.3	41.3 $\pm$ 5.7
Group IV (ABM/P-15/Hy lio)	28.8 $\pm$ 3.3	33.1 $\pm$ 2.4	116.5 $\pm$ 19.5	62.0 $\pm$ 3.0
Group V (DBM/Hy)	14.8 $\pm$ 2.9	26.9 $\pm$ 6.0	189.0 $\pm$ 56.1	42.5 $\pm$ 6.2



Graph 1. Percentages of new bone formation (% nb)-comparison between inter and intra groups at 2 weeks and 4 weeks. Group II presented the lowest percentage of bone formation, in both evaluation periods. On the contrary, at 2 weeks, Group IV presented more new bone formation than group I and III, nevertheless these differences were not statistically significant ($p > 0.05$). On the other hand, Group V presented a significant increase in new bone formation compared with group I, without any carrier ($p < 0.05$). At 4 weeks, Group IV presented a significant increase in new bone formation when compared with group I ($p < 0.05$) and there were no significant differences between groups III, V and I ($p < 0.05$). Between 2 and 4 weeks, there were statistical differences in intra-group comparison only for groups I and IV.

stabilization (Fig. 1C). This formulation also stimulated a greater cellular invasion, of osteoclastic cells, enhancing the particles resorption and the bone remodelling process, associated with a higher and better osseointegration of the particles (Boyan et al. 2006). These results suggest that Hy mainly in lyophilized formulation may increase the rate of AMB/P-15 particles resorption and bone remodelling.

As described above, sodium hyaluronate was also the carrier used in the DBM particles of group V (DBM/Hy). Regarding the results of this formulation, a substantial amount of particles only partially decalcified, were detected. At two weeks, most of the graft surface particles were coated by woven bone (Fig. 1E). Many of the demineralized particles underwent marked structural changes. Along the interface between the mineralized woven bone and demineralized matrix, spots of apparent mineral deposition were observed (Fig. 1F). These spots gradually increased in size and fused, denoting a re-calcification process (Yamashita et al. 1991; Yamashita 1992a; Yamashita 1992b; Buser et al. 1998). The trabecular new bone ($27.8 \pm 3.9\%$ and $26.9 \pm 6.0\%$ for 2 and 4 weeks) localized at the defect periphery achieved a notable maturation, that includes, however, a considerable amount of graft material. In fact, these trabeculae correspond to a composite of partially recalcified DBM allograft, woven bone, and lamellar bone.

It should be emphasized that, as found in Group IV, active bone remodelling and osteoclastic resorption were observed mainly in the partially mineralized graft particles. This is certain related to the presence the of sodium hyaluronate.

4 CONCLUSIONS

Based on the above results, it can be conclude that carriers improve handling and stiffness characteristics of the formulations tested. Additionally, it was found that carriers highly influence interparticle spacing, leading in general to a more appropriate three-dimensional distribution of the particles. Moreover, topographic localization of new bone formation closely followed the pattern of the particle distribution.

Finally, this study indicates that among the carriers tested, the sodium hyaluronate (either with the ABM/P-15 or with the DBM particles) seems to be the one that improved the most bone healing/regeneration process as well as the rate of graft resorption.

Furthermore, the lyophilized form exhibited the best performance.

ACKNOWLEDGMENTS

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An automatic morphometrics data extraction method in dental X-ray image

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ABSTRACT: Automating the process of analysis in dental x-ray images is receiving increased attention. In this process, teeth segmentation from the radiographic images and feature extraction are essential steps. In this paper, we propose an approach based on thresholding, projection profiles, mathematical morphology and Principal Component Analysis (PCA) for teeth segmentation and data extraction. Firstly, the Otsu's thresholding technique is applied. Then, mathematical morphology operators are used to improve the result. The obtained regions are searched and the larger one is taken as a reference to separate the upper and the lower jaws. We apply zonal masks in the Fourier domain to address overlapping. Teeth segmentation is finally obtained by the analysis of projection profiles. Then, we perform the feature extraction by applying the Principal Component Analysis (PCA) to get the principal axes of each tooth. Finally, we take some lengths along it that are useful for dentist diagnosis. In the experimental results we discuss the efficiency and drawbacks of the proposed pipeline.

1 INTRODUCTION

Teeth segmentation from dental x-ray films is an essential step for automating diagnosis as well as forensic procedures like postmortem identification (Said, E.H., Nassar, D.E.M., Fahmy, G. & Ammar, H. 2006). The automation of diagnosis is fundamental for computer-assisted systems in odontology while individual identification is implemented through biometric systems that use some physiological characteristics, such as fingerprints, face, hand geometry and iris (Jam, A.K. 2000).

For these applications, the segmentation step must be followed by features extraction. Dental features are manifested in root and crown morphology, teeth sizes, spacing between teeth and sinus patterns, as well as characteristics of restorations (Cameron, J.M., Sims, B.G. & Simpson, K.C. 1974).

Segmentation is a fundamental step in image analysis tasks. From a practical point of view, it is the partition of an image into multiple regions (sets of pixels) according to some criteria of homogeneity of features such as color, shape, texture and spatial relationship. These fundamental regions are disjoint sets of pixels and their union composes the original whole scene. Approaches in image segmentation can be roughly classified in: (a) Contour Based methods, like snakes and active shape models (Suri, J.S., Wilson, D. & Laxminarayan, S. 2005); (b) Region based

techniques (Zhou, J. & Abdel-Mottaleb, M. 2005); (c) Global optimization approaches (Fukunaga, K. 1990, Geman, D. & Geman, S. 1984); (d) Clustering methods, like k-means, Fuzzy C-means, Hierarchical clustering and EM (Jain, A.K., Murty, M.N. & Flynn, P.J. 1999); and (e) Thresholding methods (Sezgin, M. & Sankur, B. 2004). Among these approaches, thresholding techniques (compute a global threshold to distinguish objects from their background) are simple for implementation, with low computational cost, been effective tools to separate objects from their backgrounds. These methods have been successfully applied for document image analysis, scene processing, and quality inspection of materials. A survey of image thresholding can be found in (Kanungo, T.B., Niblack, W. & Steele, D. 1994).

There are few researches dedicated to the specific problem of dental radiograph image segmentation. In (Jain, A.K. & Chen, H. 2004), Jain and Chen worked with bitewing and panoramic dental images and apply projection histograms in order to separate the upper jaw and the lower jaw as well as to isolate each tooth from its neighbors. This is a semi-automated approach which shares some points with the method to be presented in this paper.

Besides, in (Nomir, O. & Abdel-Mottaleb, M. 2005) Nomir and Abdel-Mottaleb introduce a fully automated approach based on thresholding for teeth segmentation as well as projections to

separate each individual tooth. Image contrast enhancement and mathematical morphology has been also applied in (Zhou, J. & Abdel-Mottaleb, M. 2005).

In this paper, we focus on segmentation and feature extraction in dental x-ray images. We propose an approach based on thresholding, projection profiles, mathematical morphology and Principal Component Analysis (PCA). Thresholding is based on traditional Otsu's technique (Otsu, N. 1979). The output is processed by a close operation to improve the result. The obtained regions are searched and the larger one is taken as a reference to separate the upper and the lower jaws. A reference line is defined based on the geometry of the extracted region. We perform image rotation in order to place this line parallel to the x-axes.

Before teeth extraction, we must address the problem of overlap. We apply zonal masks in the Fourier domain to address this problem. Next, vertical projection profiles are generated and analyzed to extract each tooth following the observation that the local minima represent the limits between the teeth.

We start the feature extraction tasks, once each tooth is segmented. The tooth region is the input for the Principal Component Analysis (PCA) method (Sezgin, M. & Sankur, B. 2004) to obtain the principal axes r of the teeth. Next, we automatically determine two measures along r : the crown-body (CB) and root (R) lengths, as illustrated in Figure 1. The former is obtained through the distance from the deepest pit to the furcation. The furcation is obtained by projection profile analysis and the deepest pit is acquired by a simple search along of r direction. The latter is the distance from the furcation to the root apex. Finally, we compute the ratio CB/R.

2 PROPOSED METHOD

Our proposed method is divided in three main steps: Image Pre-Processing, Teeth Segmentation and Morphometrics Data Extraction.

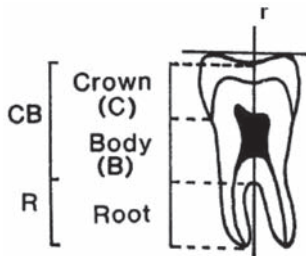


Figure 1. Teeth features: principal axes r , deepest pit (C), furcation (B) and root apex (R).

2.1 Image Pre-Processing

The dental x-ray images are obtained through the scanner, with 300 dpi resolution, in gray level scale of 8 bits.

We perform traditional average filtering in order to reduce noise and enhance object definition in the image. Then, we use the Otsu's thresholding process (Otsu, N. 1979). The result, shown in Figure 2b, is processed by a close operation in order to smooth contours and suppress small islands and holes in the binary image. The close operation is recursively applied until no changes are observed in the output image (Fig. 3a).

2.2 Teeth segmentation

After the first step, the obtained regions are labeled and the larger one is taken as the one corresponding to the separation between the jaws. This region is a reference to separate the jaws. We compute the mean path, shown in Figure 3a, between the upper and lower jaws: for each vertical $x = a$, we search for the segment inside the separation between the jaws and take the middle point of this segment. The obtained curve is the reference to separate the jaws.

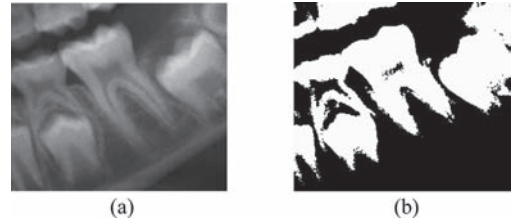


Figure 2. (a) Original image. (b) Otsu's result.

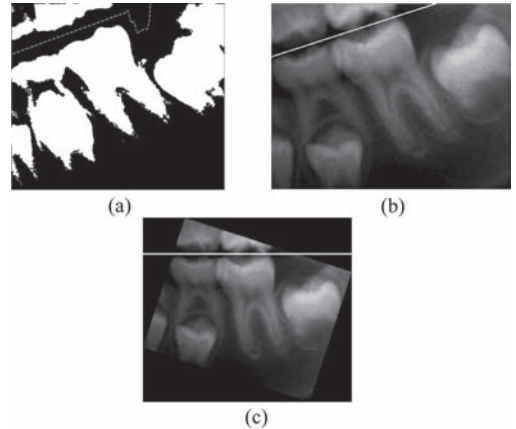


Figure 3. (a) Curve separating the upper jaw from the lower jaw. (b) Reference line for projection. (c) Original image rotated.

Now, we must compute a reference line for projection. This line is obtained taking the separation region as reference. We compute the line segment, shown on Figure 3b through connecting the first and the last point of the curve that separates the jaws. The whole original image is rotated in order to place this line parallel to the x axes, as pictured on Figure 3c.

The next step aims to separate the image in regions of interest (ROIs), each one containing a tooth. However, before this, we must address the problem of overlapping, as the one observed in Figure 4a. We apply zonal masks in the Fourier domain to address this problem. Figure 4b shows a typical mask used (band-pass filtering) and Figure 4c shows the obtained result, without overlap. The internal and external circumference's radius, in Figure 4b, are 0.01 and 1 radians, respectively. Then, we take the result (Fig. 4c) and apply a threshold of $T = 12$ for binarization (Fig. 4d).

Next, we apply vertical projection to extract each tooth. So, we project the binary image pictured on Figure 4d in the x axes. A typical projection profile is pictured on Figure 5a. We have observed that the local minima represent the limits between the

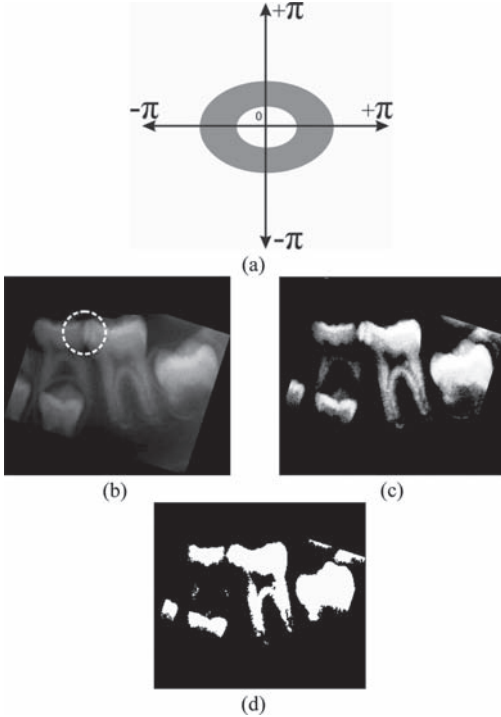


Figure 4. (a) Zonal mask in the Fourier domain. (b) Teeth overlapping. (c) Filtered image by zonal mask (overlapping elimination). (d) Binarization of the Filtered image.

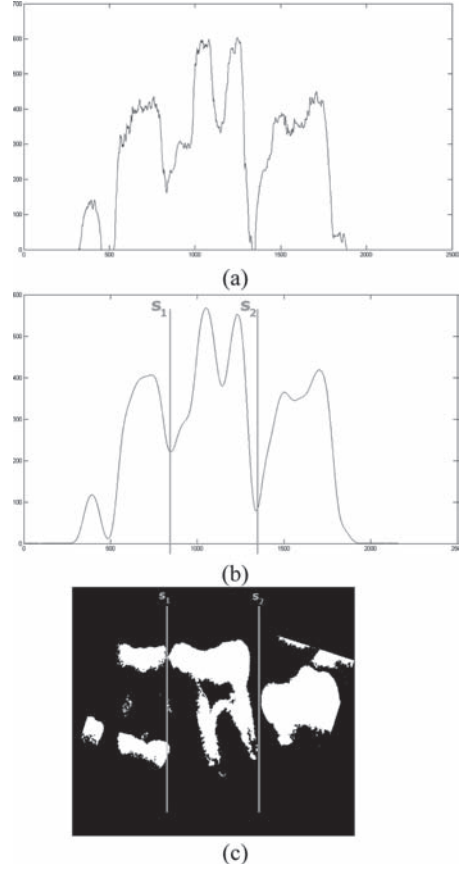


Figure 5. (a) Typical projection profile: Projection profile of the image in Figure 4. (b) Projection profile smoothed by gaussian filter. (c) Separation lines.

teeth. So, in order to reduce small scale effects, we convolve the projection profile curve with a Gaussian kernel and search for the local minima of the low-pass filtered curve. This is implemented by approximating the first derivative of the filtered curve $y(x)$ by central-differences:

$$\frac{dy}{dx}(x_i) = \frac{y(x_{i+1}) - y(x_{i-1}))}{2}$$

and searching for points that satisfies $\frac{dy}{dx}(x_i) \approx 0$.

The x-coordinates of such points define the vertical lines that separate the teeth. Finally, the projection profile of each tooth is analyzed. We observe that, in general, the low-pass filtered curve has two local maxima (root apices) and one local minimum (furcation), following the nomenclature of the scheme in Figure 1. These points allow to identify the furcation location. Teeth axes directions can be computed by the PCA.

2.3 Teeth Morphometrics Data Extraction

Figure 6a pictures the original image, rotated according to the reference line direction, with the separation lines superimposed. The pixels in the tooth region between the separation lines (Fig. 6b) are the input for the Principal Component Analysis (PCA) method (Sezgin, M. & Sankur, B. 2004) to obtain the principal axis r of the tooth.

Then, we obtain the deepest pit by searching the image along of the main axis r . The furcation is obtained by projecting the tooth in the r direction and searching for the valley between two local maxima. Next, we automatically determine two measures along r : the crown-body (CB) and root (R) lengths, as illustrated in Figure 1. The measurement of the crown-body is obtained through the distance from the deepest pit to the furcation. The latter is the distance from the furcation to the root apex. Finally, we compute the ratio CB/R. The Figure 7 shows the pipeline of our methodology.

3 EXPERIMENTAL RESULTS

The Figure 8 shows typical images that we have used in our experiments. The success of the proposed methodology is too dependent from the obtained principal axis. Besides, the reference line definition is the heart of the whole process because the separation of the teeth is very linked on this line direction.

We implement our techniques using MatLab, version 7.4. We plan to obtain a C/C++ implementation soon, in order to acquire a reliable measure of the performance. Anyway, the methodology is

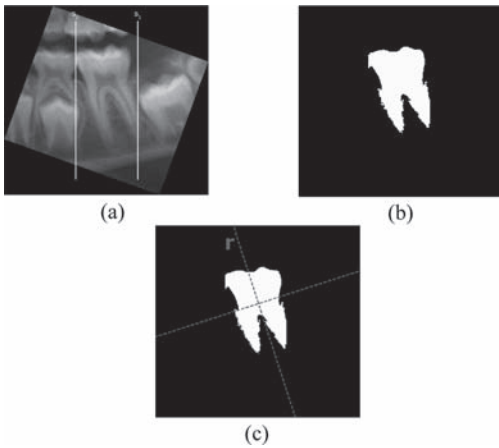


Figure 6. The PCA method in the tooth region. (a) Original image. (b) Segmented tooth. (c) PCA main axes of the segmented image.

easy to implement and it is based on optimized procedures. So, we expect a suitable performance using a C/C++ implementation.

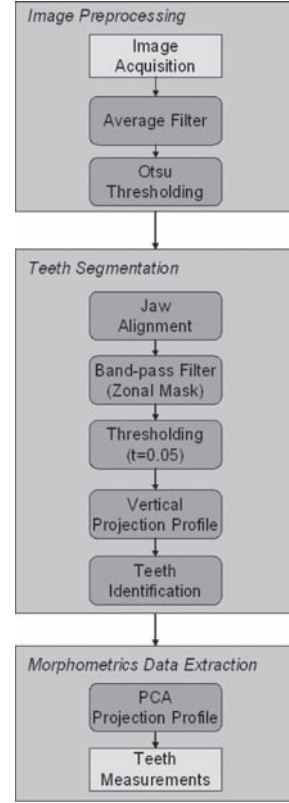


Figure 7. The pipeline of the proposed method.

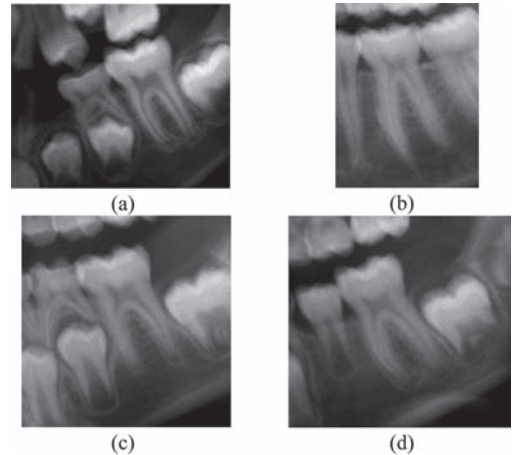


Figure 8. Test dental x-ray images used in the experiments.

Figure 9 shows the obtained principal axes for the images presented on Figure 8. The main axes of the teeth are properly calculated by PCA in these cases. However, it's very dependent from the segmentation result.

The Figure 10 shows tooth data extraction from both CB and R (Figure 9a). These features are crown-body (CB) and root (R) lengths that obtains the main axis r , where r is the ratio CB/R. This r is the morphometrics data searched in this paper.

The main drawbacks for our method are the distortions generated by the acquisition process and the difficulties for the suitable definition of the reference line.

Shape distortions due to x-ray projection need a more sophisticated algorithm, a shape model (Buchaillard, S., Ong, S., Payan, Y. & Foong, K. 2006), in order to deal with the range of possible variabilities. The reference line definition is fundamental for the projection profile as well as the ROI extraction. One possibility to improve the

efficiency of this step would be to search for the best line among a set of lines closer to the reference one.

The Zonal Mask has 0.01 and 1 radians in the Fourier domain, respectively, for the internal and external circumference's radius. The zonal mask in the Fourier domain does not harm the images without overlaps. For instance, Figure 11 shows an example without overlapping. We observe that the obtained image after zonal masking preserves the information need for the next step (projection analysis).

Besides to improve the reference line identification and distortions corrections, we also need to improve the algorithm for tooth identification. The Figure 12 shows this problem. In this case we observe that the teeth separation stage fails because parts of the teeth nearby the target one appear in the extracted region. As a consequence, the PCA result may be different from the desired one. A visual inspection indicates that the obtained principal component is not the expected tooth axis.

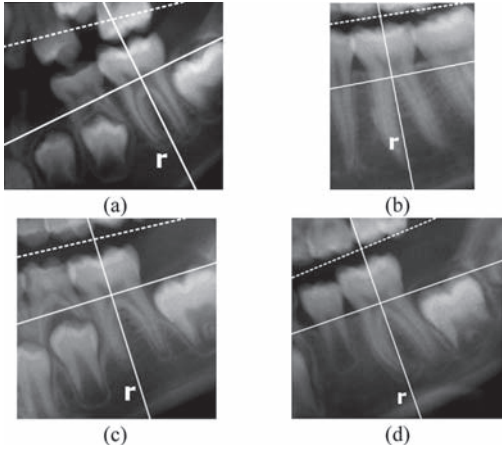


Figure 9. Original image and the obtained principal axes.

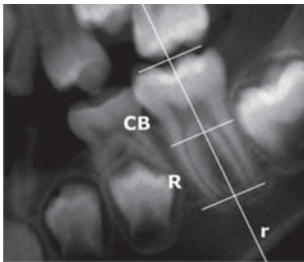


Figure 10. Automatic identification of the tooth features for the Figure 8a.

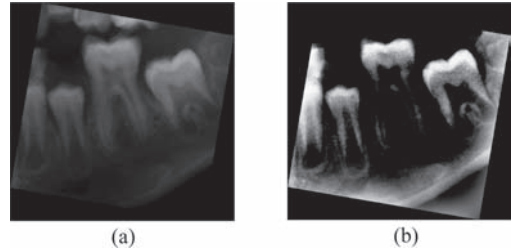


Figure 11. (a) Image without overlapping. (b) Filtered image by zonal mask.

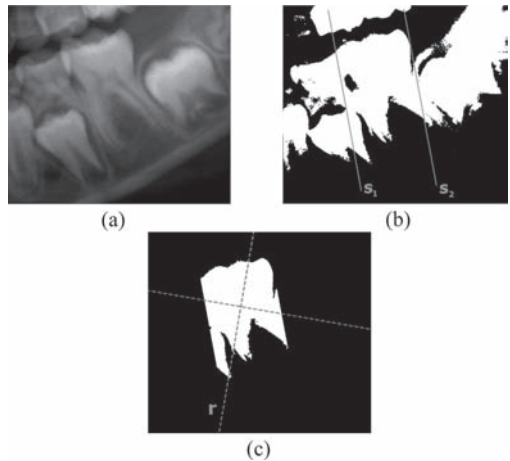


Figure 12. (a) Original image. (b) Wrong teeth segmentation and separation lines. (c) PCA result.

4 CONCLUSION

In this paper we present a method for automatic segmentation and feature extraction for dental x-ray images. The proposed methodology has been implemented using image processing tools and shape analysis (PCA) for automatically computing the principal axes of each tooth as well as the ratio CB/R.

The experimental results show that it is a promising technique, but needs improvements in the projection and PCA stages. Specifically, we must implement to search based procedure to seek for the best reference line among a set of candidate ones. Besides, a shape model for the target tooth may be designed to steer the principal axis computation.

ACKNOWLEDGEMENTS

We would like to thanks Brazilian Agencies for research support (CNPq, CAPES) and the PCI-LNCC for the funds for this work. Besides, we also thanks the graduate student Johannes Niedermaier for the valuable discussions that generates the idea of applying zonal masks in this work.

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Risk of failure at the cement-enamel junction of a human premolar tooth

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ABSTRACT: Non-carious cervical lesions (NCCL) are characterized by the loss of dental hard tissue at the cement-enamel junction (CEJ). This type of cervical tooth loss, the so-called abfraction, can be associated with cuspal flexure, due to malocclusal loads causing the tooth to laterally bend. Exceeding stresses are therefore generated in the cervical region of the tooth that cause disruption of the bonds between the hydroxyapatite crystals, leading to crack formation and eventual loss of enamel and the underlying dentine.

The aim of this study was to perform stress analyses of the first maxillary human premolar. An accurate model based on CT images of both the tooth and the periodontal ligament was employed. The three-dimensional finite element model was used to investigate the stress distributions and to compare the changes occurring between normal and malocclusion. The risk of failure at the CEJ and to crack initiation at the dentin-enamel junction was also estimated.

1 INTRODUCTION

Non-carious cervical lesions (NCCL) are characterized by the loss of dental hard tissue at the cement-enamel junction (CEJ) (Rees 2002, Levitch et al. 1994). Traditionally this has been assumed to be due to the effects of abrasion and/or erosion. More recently cervical tooth loss has been linked with cuspal flexure. It has been suggested that occlusal loads cause the tooth to flex, particularly during lateral excursion. As the tooth flexes, tensile and shear stresses are generated in the cervical region of the tooth that cause disruption of the bonds between the hydroxyapatite crystals, leading to crack formation and eventual loss of enamel and the underlying dentine (McCoy 1982, Lee & Eakle 1984). Grippo (1991) coined the term “*abfraction*” to distinguish this type of cervical tooth loss associated with cuspal flexure.

Previous finite element (Grippo 1992, Rees et al. 2003, Tanaka et al. 2003, Borcic et al. 2005) and strain-gauge studies (Owens & Gallien 1995) have found that stresses concentrated in the thin cervical enamel area, and the magnitude of these stresses exceeded the known failure stresses for enamel. Improved computer and modelling techniques render the finite element method (FEM) a very reliable and accurate approach in biomechanical applications.

The aim of this study was to develop a finite element model of the first maxillary premolar in order

to compare the stress profiles in the buccal and palatal cervical regions. The premolar was chosen because a previous study (Borcic et al. 2004) confirmed that every third premolar was affected by some form of NCCL, and a two-rooted tooth was used as model for studying the NCCL (Borcic et al., 2005). This study used a three-dimensional finite element model to investigate stress distributions and compare the changes in the stresses in normal occlusion (NO) and in malocclusion (MO).

2 MATERIALS

The mechanical properties of the enamel, dentine, pulp and periodontal ligaments are shown in Tables 1 and 2. The materials of the various

Table 1. Stiffness properties of materials.

Properties	Young's modulus (GPa)	Poisson's ratio (–)
Materials		
Enamel	80*	0.3*
Dentin	18.6*	0.31*
Pulp	0.0021 [#]	0.45 [#]
Periodontal ligament	0.0689 ⁺	0.45 ⁺

* Rees et al. (2003).

⁺ Eskitascioglu et al. (2002).

[#] Lin et al. (2001).

Table 2. Strength properties of materials.

Properties	Compressive (MPa)	Tensile (MPa)	Shear (MPa)
Materials			
Enamel	95–386 [†]	30–35 [†]	60 [†]
Dentin	249–315 [†]	40–276 [†]	12–138 [†]

[†]Zhou & Zheng (2008).

tooth structures were assumed to be isotropic, homogeneous and linearly elastic. They remained constant under the monotonically and statically applied loads.

3 METHODS

In this paper, three-dimensional finite element analyses were performed on a human intact maxillary first premolar in order to address the problems mentioned in the Introduction. An accurate finite element model based on CT images of both the tooth and the periodontal ligament has been employed. Tetrahedral elements have been used to construct the model and the contact options of full bond between periodontal ligament and maxillary bone have been used also.

The finite element model was constructed from the contours of each morphological entity (dentine, enamel, periodontal ligament and pulp) obtained from successive 541 CT-images. The CT-images were available at a spacing of 43.882 μm , thus allowing an accurate description of tooth anatomy to be obtained.

The outline of the periodontal ligament 0.3 mm wide was generated using the outline of the tooth as a guide. The dimensions of the periodontal ligament were derived from the literature (Lindhe & Karring 1989, Schroeder & Page 1990). The solid model was transferred into the FEM program Comsol 3.4 (Comsol). A three-dimensional mesh was created, and the stress distribution analysis was performed. Boundary conditions have been established on the outer surface of the surrounding ligament. It has been estimated that the boundary conditions were applied far enough from force application point to not significantly influence the stress distribution in different part of the tooth. Therefore, the ligament was clamped (all displacements fixed), thus preventing rigid body displacements in directions of all three coordinate axes. In these analyses, the contact with neighbouring teeth was not modelled by applying specialized contact elements; in fact, contact modelling would unnecessary increase the complexity of the

model without significantly influencing the final results.

Nine-noded tetrahedral elements were applied in the discretization of the tooth morphology, resulting in 152,052 elements and 29,487 nodes with 657,543 degrees of freedom. The mechanical properties of the enamel, dentine, and periodontal ligament are shown in Tables 1 and 2

In the case of normal occlusion, the three forces were applied on the occlusal surface. The forces were acted at the palatal incline of the buccal cusp, at the buccal incline of the palatal cusp, and at the palatal incline of the palatal cusp (Fig. 1a). In case of malocclusion, the force was applied to the buccal incline of the palatal cusp (Fig. 1b). During the analysis, the models of the tooth were loaded with forces N_1 , N_2 , and N_3 in the case of normal occlusion, and M in the case of malocclusion, which are assumed to be a normal chewing load. The chewing forces produced by mastication are reported to range from approximately 37 to 40% of the maximum bite force (Nakamura et al. 2001, Borcic et al. 2005). The load vectors were applied in the direction normal to the surface in order to simulate the contact with antagonistic teeth. In more detail, the force components in the coordinate system of the Figure 1 are reported in Table 3.

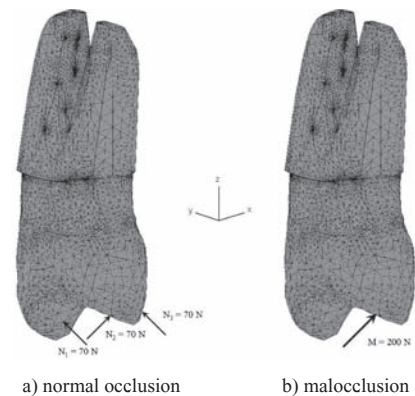


Figure 1. Loading conditions.

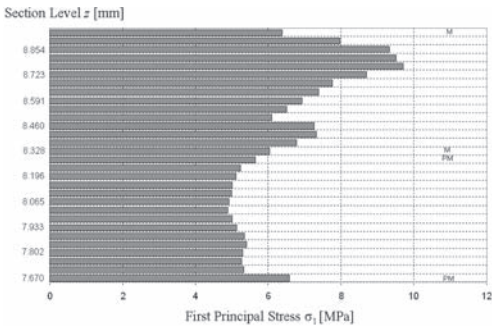
Table 3. Loading conditions.

Components		x	y	z
		N	N	N
Normal occlusion	N_1	0	49.5	49.5
	N_2	0	-49.5	49.5
	N_3	0	49.5	49.5
Malocclusion	M	0	-143	143

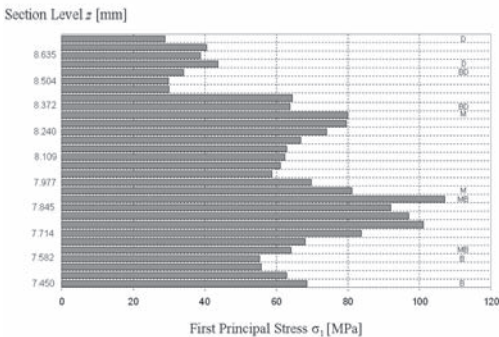
4 RESULTS

A detailed description of the stress distribution was based on 30 horizontal sections at the CEJ and on the enamel-dentin junction (EDJ), comprised between the levels $z_1 = 7.45$ mm and $z_2 = 9.030$ mm of the extremal cross-sections measured with respect to the buccal cusp. Figures 2–4 show differences in the stress distribution between the two models under different loading conditions. Two typical cases have been considered: the tooth under normal occlusion (Fig. 1a) and the tooth under malocclusion (Fig. 1b).

The results are presented as maximum and minimum principal stresses (σ_1 and σ_3 , Figs 2, 3); positive and negative values indicate that the corresponding regions are subjected to tensile or compressive stresses, respectively. Furthermore, Mises stress (σ_M) fields were analysed to estimate the risk of failure in the models (Fig. 4). In more detail, each row of the histograms in Figures 2, 3 refers to a single cross-section and indicates the sector and the tissue where the maximum/minimum value

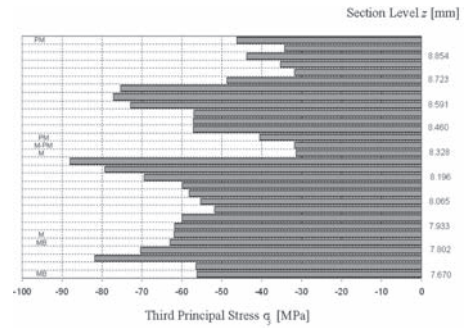


a) normal occlusion.

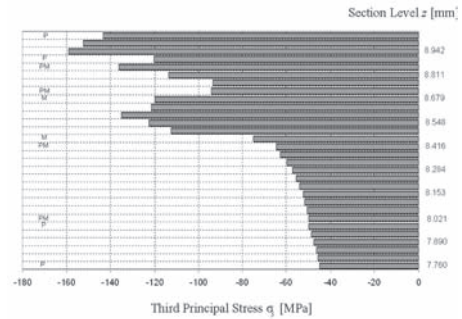


b) malocclusion.

Figure 2. Histogram of the first principal stress σ_1 [MPa].



a) normal occlusion;



b) malocclusion;

Figure 3. Histogram of the third principal stress σ_3 [MPa].

is attained by the stresses σ_1 , σ_3 , at the relevant section; each cross-section is identified by its height $z = z_1 + z_2$ on the ordinate axis; In the Figures 2, 3, the following *legenda* does hold for the sectors: B buccal, P palatal, M mesial, D distal.

Moreover, the components t_x , t_y , t_z , of the surface traction $\mathbf{t}$ on the EDJ were calculated and represented in the global x-, y-, and z-direction (Fig. 5), in order to evaluate the largest tensile and shear stresses causing possible detachment and/or slippage in normal and tangential directions with respect to the EDJ. In particular, the zones 1 and 2 in Figure 5 indicate the points where the tensile and shear stresses attained their maximum values. The tensile stress σ_n acts in direction normal to the surface of the EDJ, whereas the shear stress τ represents the effort exerted in the plane tangential to the above mentioned EDJ.

This study is devoted to predict the variations in the values of first and third principal stresses, Mises stresses, and surface traction components in the two different loading conditions, namely normal occlusion (Figs. 2a–4a) and malocclusion (Figs. 2b–4b).

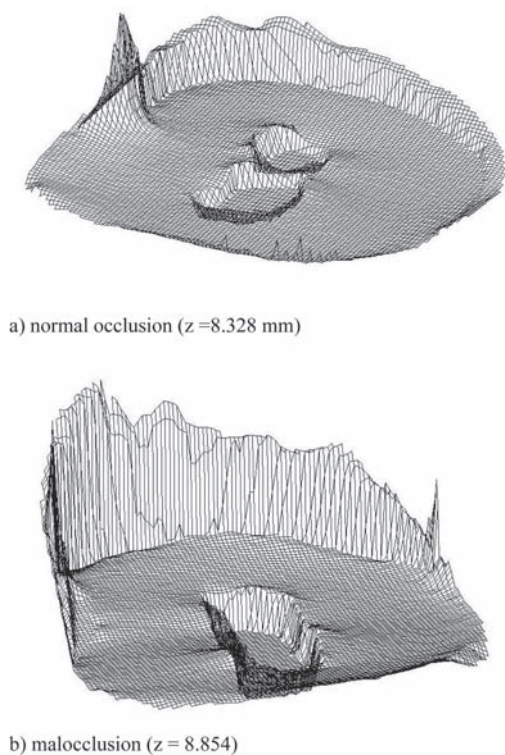


Figure 4. Distribution of the Mises stress [MPa].

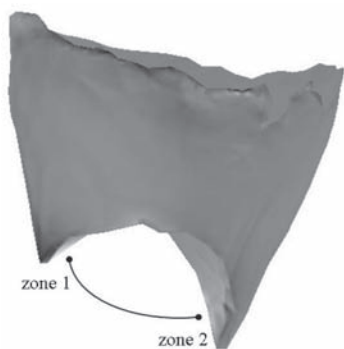


Figure 5. Interface between enamel and dentine (EDJ).

5 DISCUSSION

With reference to Figure 2a (σ_1 , NO), it can be observed that in the range from $z = 7.670$ mm to $z = 8.942$ mm the first principal stress in normal occlusion attains its larger values in the PM sector in the enamel on its outer surface; from $z = 7.802$ mm to $z = 8.284$ mm the highly stressed zones spread over the thickness of the enamel in the same sector more uniformly than in the lower

band; in the upper part from $z = 8.328$ mm to $z = 8.942$ mm, the largest values migrate toward the M sector and concentrate where the enamel gets thinner and vanishes leaving place to the dentin; the absolute maximum is reached in the M sector at $z = 8.767$ mm and is 9.7 MPa.

Figure 3a (σ_3 , NO, $z = 7.670 \div 8.942$ mm) shows that in the range $z = 7.670 \div 7.845$ mm the maximum values ($z = 7.758$ mm, $\sigma_3 = -81.8$ MPa) of the third principal stress in normal occlusion are located in the MB sector on the outer surface of the enamel; from $z = 8.416$ mm to $z = 8.942$ mm the highly stressed zones migrate toward the PM sector, passing through the M-PM sector ($z = 8.371$) and the M sector (from $z = 7.889$ mm to $z = 8.328$ mm); they concentrate where the enamel gets thinner and vanishes leaving place to the dentin; the absolute maximum is reached in the M sector at $z = 8.284$ mm and is -88.1 MPa.

The Mises stress distribution exhibits the same pattern as the third principal stress in the range $z = 7.670 \div 8.942$ mm; the absolute maximum is reached in the M sector at $z = 8.328$ mm and is 82.3 MPa. Figure 4a shows the distribution of the Mises stress at the level $z = 8.328$ mm, where the absolute maximum value is attained.

With reference to Figure 2b (σ_1 , MO, $z = 7.450 \div 8.723$ mm), it can be observed that the first principal stress exhibits its maximum value 107 MPa at $z = 7.889$ mm on the MB side in the enamel; then, the largest values of σ_1 decrease by passing from B ($z = 7.450 \div 7.582$ mm) and MB sectors ($z = 7.626 \div 7.889$ mm) to D sector ($z = 8.591 \div -8.723$ mm) through the M ($z = 7.933 \div 8.328$ mm) and BD ($z = 8.372 \div 8.548$ mm) sectors, still in the enamel.

With reference to Figure 3b (σ_3 , MO, $z = 7.760 \div 9.030$ mm), the largest values of the third principal stress are uniformly distributed in the P ($z = 7.760 \div -7.977$ mm) and PM ($z = 8.021 \div 8.416$ mm) sectors of the enamel near the interface with the dentin; then, they tend to concentrate in the P sector ($z = 8.899 \div 9.030$ mm) on the outer surface of the enamel, where they reach the maximum value -159.2 MPa at $z = 8.942$ mm. The passage through the M ($z = 8.460 \div 8.679$ mm) and PM ($z = 8.723 \div 8.854$ mm) sectors encounter two local maxima -135.1 MPa at $z = 8.591$ in the M sector and -136.1 MPa at $z = 8.854$ in the PM sector.

The Mises stress (MO, $z = 7.760 \div 9.030$ mm) gets even larger (181.0 MPa) in the PM sector ($z = 8.635 \div 9.030$ mm) at $z = 8.854$ mm; the peak values are always confined at the tip of the enamel layer of vanishing thickness; for completeness's sake, the other two local maxima 110.5 MPa at $z = 7.760$ mm and 137.3 MPa at $z = 8.591$ mm are attained respectively in the MB sector ($z = 7.760 \div 7.845$ mm), and in the M sector

($z = 7.889 \div 8.591$ mm). Figure 4b shows the distribution of the Mises stress at the level $z = 8.854$ mm, where the absolute maximum value is attained.

In the case of normal occlusion, larger compressive stresses and tensile stresses were found inside the enamel at the mesial side. In the case of malocclusion, larger compressive stresses were found at the palatal side, whereas tensile stresses were found at the mesial-buccal sector, still inside the enamel. Table 4 shows the values of the maximum and minimum principal stress values in the cervical regions for the case of NO and MO. In case NO, the peak values for the principal stress values ranged from -88.1 MPa to +9.7 MPa in the cervical areas. Case MO shows significant variation in stress values. In this case, results show increase in stress values to be reaching and tensile stress of 107.0 MPa in mesial-buccal region and compressive stresses of -159.2 MPa in the palatal region.

As far as the interface between enamel and dentin is concerned, the vector of the surface traction **t** may be a suitable measure of the stress state which could possibly initiate debonding and slippage between the two tissues. Normal stress σ_n is obtained projecting the surface traction **t** in direction **n** of the outward normal to the surface:

$$\sigma_n = \mathbf{t} \cdot \mathbf{n} \tag{1}$$

whereas shear stress τ is derived subtracting the normal stress vector $\sigma_n \mathbf{n}$ to the surface vector **t** and evaluating the intensity of the resultant vector:

$$\tau = |\mathbf{t} - \sigma_n \mathbf{n}| \tag{2}$$

If the axial stress σ_n normal to the surface and the shear stress τ tangential to the enamel-dentin interface should exceed the limits of the adhesion and cohesion strengths, σ_a , σ_c , relative motion could onset between the two tissues.

In the case of normal occlusion (Table 5), two sites seemed to be candidate: the first site is located at the zone 1 shown in Figure 5, where the components of the surface traction vector **t** in the coordinate system of Figure 1 are t_x , t_y , t_z , and the unit vector **n** of external normal has components n_x , n_y , n_z . Therefore, Eqs (1) and (2) give: $\sigma_n = -12$ MPa, and $\tau = 12$ MPa respectively. The second site is placed at

Table 4. Maximum and minimum principal stress values in the cervical region.

	σ_1 (MPa)	Sector	σ_3 (MPa)	Sector
NO	9.7	M	-88.1	M
MO	107.0	MB	-159.2	P

Table 5. Interface between enamel and dentin in NO.

		Components		
	Vector	x (MPa)	y (MPa)	z (MPa)
Zone 1	n	0.267	0.554	-0.788
	t	8	-12	10
	Normal stress (MPa)		Tangential stress (MPa)	
		-12	12	
Zone 2	n	0.771	-0.594	-0.229
	t	18.5	-9	14.5
	Normal stress (MPa)		Tangential stress (MPa)	
		16	19	

Table 6. Interface between enamel and dentin in NO.

		Components		
	Vectors	x (MPa)	y (MPa)	z (MPa)
Zone 1	n	0.267	0.554	-0.788
	t	19	-33	30
	Normal stress (MPa)		Tangential stress (MPa)	
		-36	31.5	

the zone 2 shown in Figure 5, where Eq. (1) yields $\sigma_n = 16$ MPa, and Eq. (2) yields $\tau = 19$ MPa. In the case of malocclusion (Table 6), the zone 1 appears to be the most stressed portion of the interface: $\sigma_n = -36$ MPa, $\tau = 31.5$ MPa.

The peak values of the normal and tangential stresses σ_n and τ should be compared with the corresponding strengths of adhesion and cohesion at the interface between dentin and enamel; experimental results would be welcome in order to validate the model and hence to predict possible detachments and slippages at the selected critical zones of Figure 5.

The results of the present study are in agreement with the observations in the literature (Lee et al. 2002, Tanaka et al. 2003, Lee & Eakle 1984; Heymann et al. 1991, Darendeliler et al. 1992, Borcic et al. 2005).

6 CONCLUSION

Dramatic variations were predicted in the compressive and tensile stress values in the enamel and

dentine at the cervical area in two different loading conditions (normal and malocclusion). Understanding of the cervical lesions is important for the clinical treatment and restoration of damage. This study implies the assumption that occlusal forces play a role in noncarious lesions. In the case of malocclusion, the occlusal force causes the tooth to bend by pushing against the tooth axis, and higher tensile stresses are produced on the cervical region. Enamel and dentin have been modelled as isotropic and not as orthotropic materials. The FEM model represented a static situation; indeed the loading of the structure exhibits dynamic and cyclic characteristics.

However it is apparent from the above results that the response of the structure is different if asymmetrical loading is considered. Significant weakening in the continuity of the structure of the hard dental tissues causes the increase of the stresses in the cervical region. The results of this research provide an outlook in the biomechanical aspects related to the clinical developments of NCCLs.

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A mechanobiological model for bone ingrowth on dental implants

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ABSTRACT: The main contribution of this work is the proposal of a new mathematical framework based on a set of reaction-diffusion equations that try to model the main biological interactions occurring at the surface of implants and is able to reproduce several biological features of the osseointegration phenomenon. Numerical simulations reveal that the model can reproduce the differences between contact and distance osteogenesis depending upon the specific surface microtopography and the influence of the geometry of the threads of the implant.

1 INTRODUCTION

The number of different types of bone implants and prostheses that are currently being used in clinical practice is great and follows an increasing trend. However, the convincing clinical achievements, which are shared in general by the majority of bone implants, are not correlated with an equivalent degree of understanding of the basic mechanisms of peri-implant bone healing.

Thus, the goal of this contribution is to propose a new biological model for the study of bone ingrowth on endosseous implants that, by means of considering the main biological interactions occurring between the tissue and implant, is able to reproduce several significant features of the osseointegration phenomenon. Unlike other previous models, ours is focused on the early stages of bone healing, taking into account immediate events upon implantation of biomaterials such as platelet activation, what allows reproducing in a simple way the effect of surface microtopography. Slower long-term processes, i.e. bone remodelling, are here contemplated only in a simplified manner.

This makes sense if one considers, first, that much effort has already been devoted to the mathematical modelling of bone remodeling; second, that it is reported that in many types of bone implants most failures occur within the first year and implant loss is significantly lower in subsequent years (see, for example, Goodacre et al. (1999) for the case of single unit dental implant restorations). This indicates that the critical issue for this type of implants is the early bone healing and represents

a difference with other types of implants such as cemented hip prostheses, in which long-term failure due to bone resorption caused by stress shielding or crack growth within the cement mantle is a typical phenomenon.

In order to validate the model, different computational simulations using the finite element method have been performed. In the first set of simulations, we have focused on the study of the influence of the implant surface microtopography, which is known to be one of the key factors affecting the biological performance of bone implants. In the second set of simulations, attention has been paid to the influence of the geometry of the implant threads on bone deposition. The results are in the line of what the experiments dictate and show that model is successful in reproducing the influence of the analysed factors.

2 MATHEMATICAL MODEL

2.1 General framework

We adopt a continuum approach and, consequently, are interested in the spatio-temporal evolution of the volumetric concentration of each specie. Therefore, our model is based upon the fundamental conservation law for the concentration of each specie $Q = Q(\mathbf{x}, t)$ at time t and spatial position $\mathbf{x}$:

$$\frac{\partial Q}{\partial t} = -\nabla \cdot \mathbf{J}_Q + f_Q \quad (1)$$

where $\mathbf{J}_Q$ is the flux (rate of outgoing matter per unit area) of specie Q and f_Q the rate of net production of Q .

The different variables of the model can be classified into three groups: cell densities, growth factor concentrations and matrix volume fractions. We have considered three different types of cells: platelets, osteogenic cells (mesenchymal stem cells) and osteoblasts, whose respective densities will be denoted by c , m and b . The explicit inclusion of platelets is new in this kind of applications and, as will be shown later, is necessary to account for the early stages of bone healing. Note that c stands for the density of the total population of platelets, activated and not.

A wide diversity of growth factors and other signalling molecules are known to play a role in the mediation of bone healing mechanisms, but a model in which a large number of them were individually taken into account, as well as being extremely complex and confusing, would be completely impractical due to the lack of quantitative experimental results needed to fit the parameters. This is why we have preferred to gather all the growth factors into two generic types: the first one, with concentration denoted by s_1 , corresponds to the release of activated platelets (PDGF, TGF- β), whereas the second one, s_2 , represents the set of signalling molecules secreted by osteogenic cells and osteoblasts (BMPs, TGF- β superfamily).

The extracellular matrix can be composed of three different structures: first, the fibrin network, whose volume fraction is denoted by v_f , that is assumed to be the only starting component since we do not address the problem of coagulation; second, woven or immature bone, v_w , that is laid down by osteoblasts and, third, lamellar or mature bone, v_b , that comes from remodelling of woven bone. Finally, the concentration of adsorbed proteins, p , appears in the equations but is not a model variable since its value is assumed to be known a priori as a function of the microtopography of the implant surface. It reaches its maximum value at the surface of the implant and decreases very fast as we move away from it, taking value zero in the rest of the domain.

In next sections we particularise the evolution of every specie of the model.

2.2 Platelets, $c(\mathbf{x}, t)$

$$\frac{\partial c}{\partial t} = \nabla \cdot [D_c \nabla c - H_c c \nabla p] - A_c c \quad (2)$$

The contribution to the cell flux is random dispersal. Thus it has been here modelled as a first approximation by a linear diffusion term with coefficient D_c , and the cell adhesion to the implant surface.

Platelet adhesion has been found experimentally to depend on the microtopography of the surface that alters the concentration of adsorbed proteins to the implant surface. Therefore, it has been modelled as a linear “taxis” term, depending on the gradient of p with coefficient H_c . A high platelet concentration is assumed at the beginning and thus the only kinetic term comes from cell removal due to inflammatory mechanisms with linear rate A_c .

2.3 Osteogenic cells, $m(\mathbf{x}, t)$

$$\begin{aligned} \frac{\partial m}{\partial t} = & \nabla \cdot [D_m \nabla m - m(B_{m1} \nabla s_1 + B_{m2} \nabla s_2)] \\ & + \left(\alpha_{m0} + \frac{\alpha_{ms1}}{\beta_m + s_1} + \frac{\alpha_{ms2}}{\beta_m + s_2} \right) m \left(1 - \frac{m}{N} \right) \\ & - \frac{\alpha_{mb}s_1}{\beta_{mb} + s_1} m - A_m m \end{aligned} \quad (3)$$

Osteogenic cell flux comes from random cell movement that can be biased by the presence of growth factors. Mathematically this is modelled by means of linear diffusion, with coefficient D_m , and linear chemotaxis along gradients of the growth factors s_1 and s_2 with coefficients B_{m1} and B_{m2} . For the kinetics, there is a proliferative term consisting of a logistic growth with a natural linear rate α_{m0} that can be enhanced by the presence of s_1 and s_2 ; phenotypic differentiation into osteoblasts is stimulated by the growth factor s_1 ; and natural cell death is assumed to be produced with a linear rate A_m .

2.4 Osteoblasts, $b(\mathbf{x}, t)$

$$\frac{\partial b}{\partial t} = \frac{\alpha_{bs1}}{\beta_b + s_1} m - A_b b \quad (4)$$

Osteoblasts remain on the surface of the bone matrix they secrete and therefore we can assume that there is no flux of this cellular type. The kinetics has a source term of differentiation from the osteogenic phenotype and a decay term representing differentiation into osteocytes A_b .

2.5 Generic growth factor 1, $s_1(\mathbf{x}, t)$

$$\frac{\partial s_1}{\partial t} = \nabla \cdot [D_{s1} \nabla s_1] + \left(\frac{\alpha_{c1}p}{\beta_{c1} + p} + \frac{\alpha_{c2}s_1}{\beta_{c2} + s_1} \right) c - A_{s1}s_1 \quad (5)$$

Random dispersal of the growth factor is modelled with a linear diffusion term with coefficient D_{s1} . The first kinetic term takes into account the secretion of s_1 by platelets that depends on the degree of activation. Platelet activation in turn is assumed to be fostered by the concentration of adsorbed proteins

p and the own growth factor s_1 . There is also a natural decay of the growth factor with rate A_{s1} .

2.6 Generic growth factor 1, $s_2(\mathbf{x}, t)$

$$\frac{\partial s_2}{\partial t} = \nabla \cdot [D_{s2} \nabla s_2] + \frac{\alpha_{m2}s_2}{\beta_{m2} + s_2} m + \frac{\alpha_{b2}s_2}{\beta_{b2} + s_2} b - A_{s2}s_2 \quad (6)$$

The structure of the equation is completely equivalent to the one of s_1 , although in this case there are two source terms corresponding to secretion of s_2 by osteogenic cells and osteoblasts. This secretion is enhanced by the own growth factor s_2 .

2.7 Fibrin network volume fraction, $v_f(\mathbf{x}, t)$

$$\frac{\partial v_f}{\partial t} = -\frac{\alpha_w s_2}{\beta_w + s_2} b v_f (1 - v_w) \quad (7)$$

Initially the whole volume between bone and implant is assumed to be filled with a fibrin network and v_f takes value 1. Therefore the only kinetic term comes from partial substitution of the fibrin network by woven bone matrix, that is, secretion of new bone by osteoblasts, stimulated by s_2 . Additionally, it is reasonable to consider that the rate of bone secretion takes its maximum value when there is no formed bone ($v_f = 1$) and decreases as the fibrin network is substituted progressively by bone, until reaching a null value when $v_w = 1$. Consequently, the rate of fibrin adsorption is zero when the fibrin concentration is zero, which is logical and necessary to avoid the prediction of unrealistic negative concentrations.

2.8 Woven bone volume fraction, $v_w(\mathbf{x}, t)$

$$\frac{\partial v_w}{\partial t} = \frac{\alpha_w s_2}{\beta_w + s_2} b v_f (1 - v_w) - \gamma v_w (1 - v_l) \quad (8)$$

Woven bone formation is taken into account by the first term, which was explained in the previous paragraph and describes the formation of woven bone that replaces the initial fibrin network. Note that the secretion of woven bone is stopped when its volume fraction reaches the maximum value, 1.0, avoiding the appearance of unrealistic volume fractions larger than 1.0. The second term is a very simple way of considering remodelling of woven bone into lamellar bone.

2.9 Lamellar bone volume fraction, $v_l(\mathbf{x}, t)$

$$\frac{\partial v_l}{\partial t} = \gamma v_w (1 - v_l) \quad (9)$$

Finally, the only contribution to the evolution of lamellar bone comes from remodelling of woven bone. Note also that bone remodelling stops when the totality of woven bone has turned into lamellar bone, as wanted.

3 NUMERICAL SIMULATION: BONE INGROWTH AROUND A DENTAL IMPLANT

3.1 Description of the simulation

As a first approach, we propose a simple two dimensional simulation in a domain that reproduces the cavity between the host bone and two threads of a screw-shaped dental implant (Figure 1). In particular, we have chosen a typical implant geometry characterised by threads with inclined walls. The equations have been solved with two different levels of concentration of adsorbed proteins: $p = 0.5 \mu\text{g}/\text{mm}^2$ at the surface of the implant, simulating the case of a high microtopography surface implant, and $0.1 \mu\text{g}/\text{mm}^2$, in the case of an implant of low surface microtopography. These values fall within the range of experimental measurements of plasma proteins adsorption on titanium surfaces (Sela et al. 2007) and were imposed at the nodes belonging to the surface of the implant. Moreover, p was assumed to decrease linearly with the distance to the implant surface, reaching value zero at a distance of 0.1 mm. In the rest of the domain situated farther than 0.1 mm, p took value zero. The reason to simulate with two different values of protein adsorption lies in the fact that we want to show the ability of the model to reproduce one of the most relevant features of peri-implant bone healing, which is the difference between contact and distance osteogenesis depending on the implant surface properties, a matter of the utmost importance

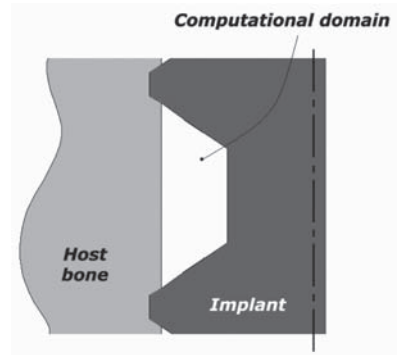


Figure 1. Sketch of the insertion of a screw-shaped implant in a drilled cavity of bone, where the computational domain is highlighted.

in the design of new models of implants. We recall here that rough implants tend to enhance the formation of bone on the surface of the implant, as opposed to polished implants, where the front of ossification typically travels from the host bone towards the implant (Berglundh et al. 2003).

As initial conditions, we have considered a concentration of 2.5×10^8 platelets/ml, being this high value characteristic of blood, and residual concentrations of osteogenic cells and osteoblasts of 10^3 cells/ml. Both growth factors are also present at the initial time at a very low concentration of 1 ng/ml. Zero flux boundary conditions have been applied for all the species of model at the surface of the implant and the host bone, except for the concentration of osteogenic cells, that was fixed to 2×10^3 cells/ml at the surface of bone during the first 14 days. The healing period that was simulated consisted of twelve months, more than sufficient in clinical practice to obtain a full osseointegration of the implant and achieve a high degree of remodeling of woven bone into lamellar bone.

The full model has been numerically solved by means of the finite element method, using second order spatial interpolation with eight-node Serendip elements, a generalised trapezoidal method for the integration of temporal derivatives and a Newton-Raphson implicit scheme with full linearisation of the residual vector for the treatment of the nonlinearities. This formulation has been implemented in an UEL subroutine of the commercial software ABAQUS 6.6. The model parameter values were estimated from the literature as discussed in detail in Moreo et al., in press.

3.2 Results

We must highlight in first place the substantial difference early found in the density of platelets c and the concentration of the growth factor s_1 depending on the microtopography. In Figure 2 it can be observed that only one day after implantation the high concentration of adsorbed proteins on the surface of the high microtopography implant leads to an increase in the number of platelets near the surface, compared to the situation of a low microtopography implant, where the density of platelets is almost uniform. Actually, an almost 4-fold variation in the concentration of platelets at the very surface of the implant was obtained between the two types of surfaces, what corresponds with the range of experimental observations (Kikuchi et al. 2005). This alteration of platelet concentration was expected, since H_c was precisely chosen in such a way that such an increase took place when the implant surface microtopography, i.e. the concentration

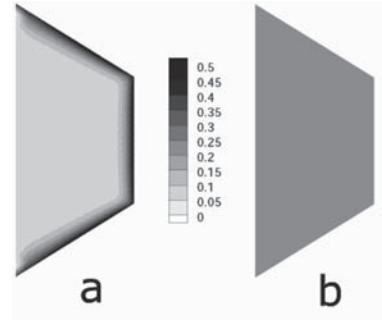


Figure 2. Density of platelets c ($\times 10^9$ cells/ml) one day after placement of the implant in the case of an implant with (a) high microtopography and (b) low microtopography.

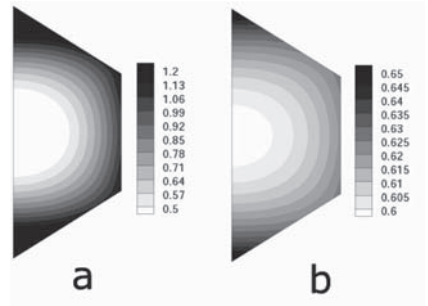


Figure 3. Concentrations of growth factor s_1 ($\times 100$ ng/ml) after 14 days in the case of implant with (a) high microtopography and (b) low microtopography.

of adsorbed proteins, was changed from low to high. This increase in platelet concentration and activation leads to not only a higher concentration of s_1 after 14 days of healing, but also to a gradient of this growth factor, being its concentration markedly higher near the implant surface and decreasing as we move away (see Figure 3a). On the other hand, a low surface microtopography does not favour the formation of this gradient, as we appreciate in Figure 3b, where a variation of only 10% is obtained along the whole cavity.

It is evident that this early discrepancy between the two types of implants will strongly condition the subsequent healing phases. For example, in Figures 4 and 5 the temporal evolution of the volume fraction of lamellar bone has been depicted. Significant differences appear between the two microtopographies. In the case of the rough implant (Figure 4), the model predicts the formation of new bone from the surface of the implant towards the inside of the cavity, which is commonly denoted as contact osteogenesis. This phenomenon is especially clear during the first month after implantation, since it is in

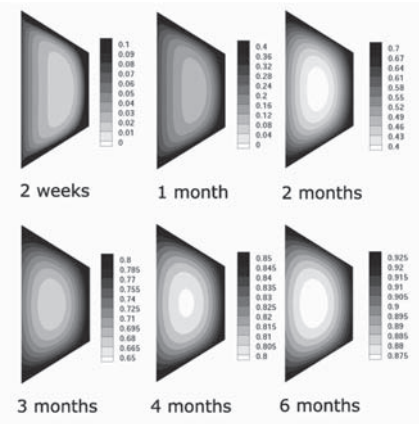


Figure 4. Temporal evolution of the volume fraction of lamellar bone around an implant with a high surface microtopography.

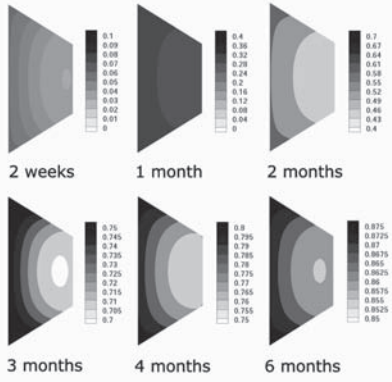


Figure 5. Temporal evolution of the volume fraction of lamellar bone around an implant with a low surface microtopography.

this period when a more pronounced gradient in the volume fraction of lamellar bone develops. An ossification front also propagates from the surface of the peri-implant bone directed to the implant surface, since the surface of bone is in any case an important source of osteogenic cells. However, this second mechanism leads to a more constant and slow bone ingrowth. We could say that contact osteogenesis is responsible for the very early formation of new bone on the surface of the implant, what ensures a fast mechanical stability and prevents at the same time the formation of a fibrous capsule, but it is the host bone, acting as a source of osteogenic cells, that guarantees the full ossification of the chamber between bone and implant by means of distance osteogenesis. The coexistence of these two ossification

mechanisms was observed experimentally in a rough SLA implant surface by Berglundh et al. (2003) and is evident in our simulations from the observation of the results of Figure 4 after one and two months of healing.

In the case of the low microtopography implant, the features of the ossification process are noticeably different. In first place, the mechanism that inevitably prevails is distance osteogenesis, as shown in Figure 5 after healing periods of two weeks and one month. Observe that the rate of bone formation during this initial phase is appreciably slower than in the previous situation. Actually, the volume fraction of lamellar bone after one month reaches a value of 0.3 at the surface of bone, whereas with the rough implant values of 0.4 were already obtained along the whole surface of the implant at that very moment, what is in agreement with experimental observations of (Puleo & Nanci 1999), where a difference of 30% in the rate of bone formation was found between contact and distance osteogenesis. Another interesting issue is the fact that the fraction of bone is now quite uniform along the chamber, what can be explained by the absence of contact osteogenesis, that clearly leads to a more uneven distribution. Finally, the global levels of lamellar bone volume fraction achieved after four, six and twelve months are in general terms comparable to the obtained with the high microtopography implant, although smaller in all cases by a 5–10% amount.

4 NUMERICAL SIMULATION: BONE INGROWTH AROUND A GROOVED DENTAL IMPLANT

4.1 Description of the simulation

In this second simulation, we used the model to study the effect of the geometry of threads of the implant. In particular, we considered a screw-shaped implant characterized by threads with inclined walls and longitudinal grooves of circular section, proposed by some manufacturers as an innovative way to enhance the formation of new bone and achieve a better interlock between bone and implant (Hall et al. 2005). The model parameters and the initial and boundary conditions that were used are the same that those of Section 3.

4.2 Results

Firstly, we note that the previous differences between contact and distance osteogenesis also appear now when different surface microtopographies are tested.

However, more important that this result, already detected in Section 3, is the effect of the geometry of the grooves on the formation of new bone. From Figures 6 and 7 we can conclude that the grooves promote early deposition of bone inside them and hence have a stimulating effect on bone formation. This fact has been observed experimentally (Hall et al. 2005) and, according to our model, is a direct consequence of curvature of the implant surface at the grooves, which leads to an accumulation of s_1 inside them. This provokes that osteogenic cells migrate preferentially towards the grooves, where, besides, the rate of proliferation and differentiation is enhanced due to the high concentration of s_1 . As expected, the stimulating effect of the grooves on peri-implant bone

formation is more pronounced in the case of the rough implant, given the large concentration of proteins adsorbed at the surface of the grooves.

5 CONCLUSIONS

In this work we have proposed a new mathematical model for peri-implant bone formation that incorporates several novel elements over the current literature. Firstly, it is focused on the early stages of bone healing and considers platelet activation. This permits to take into account the influence of the surface microtopography on the biology of peri-implant bone formation. This involves an advance, since the majority of theoretical works existing in the literature have neglected this influence. As a consequence, our model successfully reproduces the differences between contact and distance osteogenesis as a function of the implant roughness. In addition, the model also predicts the early bone formation inside grooves placed at the surface of threads, providing a plausible explanation for this phenomenon.

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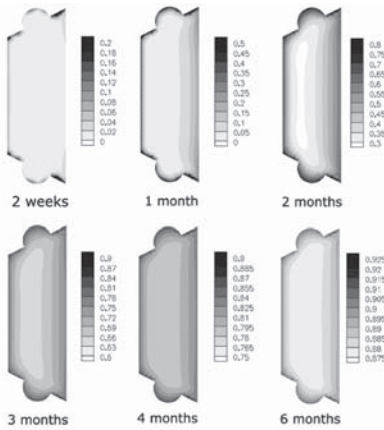


Figure 6. Temporal evolution of the volume fraction of lamellar bone around an implant with a high surface microtopography.

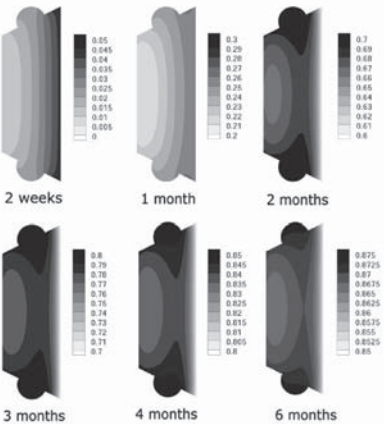


Figure 7. Temporal evolution of the volume fraction of lamellar bone around an implant with a low surface microtopography.

Optical behavior of two dental bleaching agents irradiated with different wavelength

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ABSTRACT: The optical behavior of two dental bleaching agents were evaluated when irradiated with different wavelengths in the visible spectrum, the Whiteness Hp and Whiteness Hp Maxx gels from FGM Company. The agents were irradiated by LEDs (light emitting diode) with wavelength of 470 nm (blue) and 568 nm (green). For the optical analysis a LDR (light dependent resistance) photo sensor was used to measure the intensity of the light transmission through the bleaching gel. The photo sensor is part of a benching test which was developed for this kind of analysis and it is made by a LDR connected to a digital multimeter. The testing bench has shown itself efficient to analyze the transmission of different wavelengths through the Whiteness Hp and Whiteness Hp Maxx gels. The three factors analyzed in the experiment: gel, wavelength and time, have a significant Influence in the light transmission into the gel.

1 INTRODUCTION

In dentistry, blue LEDs have been used to photo activate composites resins and to accelerate the oxyreduction reaction of dental bleachers (Christensen 2003, Stahl et al. 2000, Tarle et al. 2002, Knezevic et al. 2002). Recently, some companies have released in the market equipments with green LEDs, justifying that those were absorbed by the pigments inside dental bleachers. The objective of this experiment is to analyze the transmission of the blue and green lights inside the Whiteness Hp (FGM—Brazil) and Whiteness Hp Maxx (FGM—Brazil) gels.

References about the utilization of optical systems have analyzed the penetration of laser beams in organic tissues (Kolari & Airaksinen 1993, Kolarova et al. 1999, Lucas et al. 2002). However, it hasn't been found the utilization of one optical sensor to analyze the incidence of light beams in dental bleachers.

2 METHODOLOGY

The LEDs used were 5 mm diameter, 5 V tension and 50 mA current. The blue LED operated in 470 nm and the green in 568 nm.

Resistance controllers were utilized in order to obtain the same radiance flow from the LEDs and their entrance tension could be modulated. The testing bench was set in a contact matrix (protoboard), under a source of 12 V. At the beginning the radiance flow of all LEDs was measured

with the minimum resistance of the circuit. After obtaining the highest flow possible in each LED, an intensity modulation was taken with each LED emitting 0,8 mW. The lower intensity is due to the fact that there are no green LED with high radiance flow.

The measurement of the radiance flow or the light potency of the LEDs was done by a photodiode (PD300—OPHIR—Israel). Before each experiment a new calibration took place to check if all LEDs emitted 0,8 W of light energy

The results shown in this study were based on the utilization of a hydrogen peroxides 35% bleacher, Whiteness HP and Whiteness HP Maxx by FGM.

Both bleachers have a purple pigment with an absorption peak in 521 nm, the difference between them is a yellow and a blue pigment in the Whiteness Hp Maxx gel, those together turn up green. Besides that, silica crystals are present in the Hp Maxx gel.

During the bleaching process, the Whiteness Hp, which is purple at the beginning, will end up transparent (Fig. 1).

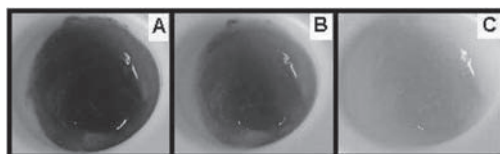


Figure 1. Chemical reaction process of Whiteness Hp.

The Whiteness HP Maxx product is purple at the beginning of the reaction turns green during the bleaching process (Fig. 2).

A chamber, composed by a photo sensor called LDR (Light Dependent Resistor) was utilized to analyze the light transmission from inside the bleacher gel placed in between the LDR and the LED.

The LDR showed a linear response to the spectral line used in the experiment (470 nm to 570 nm).

Before each reading, the system was calibrated in order to have no distortions.

A time set of 1 minute was established to manipulate the gel, place it inside the tests chamber and turn the LEDs on. Since the gels used needed to be manipulated, 9 drops of peroxide to 3 drops of thickener was established, 30 seconds to manipulate and 30 seconds to close the tests chamber.

The transmission profile of the bleacher gel towards each LED emission was observed by a period of five minutes. To each wavelength five readings of transmission were taken.

3 RESULTS

The Figure 3 shows the light transmission profile of the blue light with the wavelength of 470 nm in the bleaching gels Whiteness Hp and Whiteness Hp Maxx. At the first 90 seconds the behavior of both gels was similar. After that period the transmission of the blue light in Whiteness Hp Maxx gel was much lower than Whiteness Hp gel. This probably happened due to the dissociation of the purple dye

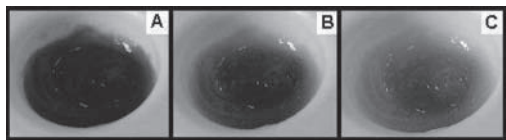


Figure 2. Chemical reaction process of whiteness Hp Maxx.

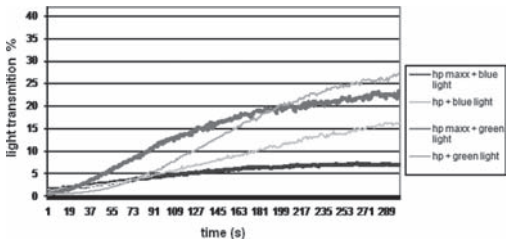


Figure 3. Transmission of blue and green light through Whiteness Hp and Whiteness Hp Maxx gels.

Table 1. Described measurements of the variable light transmission considering the gel, wavelength, time and follow up.

Gel	Wavelength	Time	Described Measures				
			Minimum	Maximum	Average	D.p.	CV
Hp	BLUE	1 seg.	1,517	1,517	1,517	0,000	0,00
Hp		1 min	2,759	3,310	3,034	0,218	7,19
Hp		2 min.	6,621	7,034	6,759	0,169	2,50
Hp		3 min.	9,931	10,207	10,069	0,098	0,97
Hp		4 min.	13,379	13,793	13,517	0,169	1,25
Hp		5 min.	16,000	16,552	16,276	0,218	1,34
Hp	GREEN	1 seg.	0,522	0,617	0,569	0,034	5,98
Hp		1 min	2,230	2,514	2,324	0,116	4,99
Hp		2 min.	9,488	9,772	9,677	0,111	1,15
Hp		3 min.	17,505	17,647	17,552	0,058	0,33
Hp		4 min.	23,577	23,861	23,719	0,121	0,51
Hp		5 min.	26,565	27,040	26,898	0,190	0,71
Hp Maxx	BLUE	1 seg.	1,379	1,655	1,517	0,138	9,10
Hp Maxx		1 min	3,034	3,310	3,172	0,138	4,35
Hp Maxx		2 min.	4,828	5,517	5,103	0,276	5,41
Hp Maxx		3 min.	5,931	6,621	6,207	0,258	4,16
Hp Maxx		4 min.	6,621	7,172	6,897	0,218	3,16
Hp Maxx		5 min.	6,759	7,310	7,034	0,195	2,77
Hp Maxx	GREEN	1 seg.	0,474	0,569	0,522	0,034	6,51
Hp Maxx		1 min	6,309	6,594	6,499	0,111	1,71
Hp Maxx		2 min.	13,520	13,994	13,852	0,193	1,39
Hp Maxx		3 min.	18,121	19,070	18,833	0,403	2,14
Hp Maxx		4 min.	20,114	20,588	20,398	0,177	0,87
Hp Maxx		5 min.	22,486	23,008	22,818	0,198	0,87

inside Whiteness Hp gel, which happens during the oxidation of the oxygen peroxide. Besides having the purple dye, Whiteness Hp Maxx also has blue and yellow both keep absorbing the blue light, however with a lower efficiency than the purple one.

In the same figure there is the profile of green light transmission with wavelengths of 568 nm in the bleachers gels Whiteness Hp and Whiteness Hp Maxx.

The LEDs with blue wavelength (470 nm) were the ones which showed the lower transmission score through the gels Whiteness Hp and Whiteness Hp Maxx when the period of five minutes of irradiation was observed.

The transmission into the gel Whiteness Hp Maxx (maximum of 7,31%) under similar conditions of time and irradiation, has shown statically significant comparing to Whiteness HP (maximum 16,55%).

Likewise, when the green wavelength LED was used, the transmission in the Whiteness HP Maxx (maximum of 23,00%) under similar conditions of time and irradiation has shown statically significant comparing to Whiteness HP (maximum of 27,04%).

4 DISCUSSION

The methodology used has shown efficient to determinate the transmission of light in a gel where a dynamic chemical process took place. The results showed that there was a significant statistical difference when changes such as the wavelength, the time and the pigments inside the dental bleacher occurred.

The spectrum of a determined photoinitiator in a gel of dental bleaching shows an absorption peak and an absorption band, however, many photoinitiators lose their optical properties when reacting with hydrogen peroxide and make a new compound. This occurs with the purple pigment of Whiteness HP and Whiteness HP Maxx, they have an absorption peak in 521 nm, but during the reaction this photoinitiator is broken and the gel no longer absorbs at this peak.

Luk et al. (2004) suggested that dentists who use the bleaching technique in their practices, with the help of a light source, should take into consideration the specificity of the bleaching gel towards the source.

Most of the commercial equipments for dental bleaching have blue light emitting LEDs between 450 nm and 480 nm. According to the Beer's Law it would be interesting that those bleachers have a photo absorbed pigment in the same spectral band.

The existence of such pigment would promote a higher absorption of the emitting radiation, increasing the gel temperature and lowering the transmission of radiation to the dental pulp.

The hydrogen peroxide concentration in the bleachers Whiteness Hp and Whiteness Hp Maxx could change according to the number of drops manipulated, the size of the dispenser, the time and the way how the gel is manipulated amongst many others. Due to the fact that the concentration was not pre-determined for the application, the increase or decrease of it could create variations in the speed of the reaction and its final balance. An increase in the concentration of peroxide, such as the increase of temperature would increase the speed of the reaction, creating a larger amount of the product (free radicals) in a shorter time period. The increase of the concentration of peroxide would also increase the initial enthalpy increasing the liberation of heat during the reaction.

According to Freedman (1990) the hydrogen peroxide has a low molecular weight. This characteristic contributes to the diffusion of the substance through enamel and dentin, reaching the dental pulp. The pigments inside the deeper layers are usually more difficult to bleach, demanding a higher bleaching time. Therefore, the amount of hydrogen peroxide inside the pulp chamber shouldn't be excessive which according to Bowles & Thompson (1986) causes the inhibition of the production of enzymes responsible for the normal function of the dental pulp.

It was certain that Whiteness Hp Maxx absorbs more visible wavelengths than Whiteness Hp, specially blue and green. In the study of Wetter et al. (2004) Whiteness Hp was shown to be an efficient bleacher comparing to Opalescence Xtra (Ultradent—USA), the authors of this

study compared a laser diode action and a LED equipment over those bleachers and the bleaching results were significantly better with the combination laser and Whiteness Hp. However, it would be interesting if the purple pigment inside both Whiteness Hp and Whiteness Hp Maxx wouldn't suffer a color change during the bleaching process, hence it is the major responsible for the light absorption of blue and green light inside the gels. The decomposition of the purple pigment makes Whiteness Hp gel transparent, allowing a higher light transmission in it. This transformation in the product would turn it safer to possible higher temperatures changes.

These data are all relevant, since the blue wavelength is the most used in dentistry for dental bleaching. Taking under consideration the increase of radiation to the pulp, by laser, LED or a conventional halogen light curing unit, there is an increase in the intrapulpal temperature and an ideal combination of wavelength, time and dye could lower these possibilities. Clinically, the decrease of light transmission over the bleaching gel would lower the undesirable effects such as tooth sensibility, which occurs after and during the bleaching process making the bleaching technique safer for the dental professional point of view and more comfortable for the patient.

5 CONCLUSIONS

The testing bench showed efficient to analyze the transmission of different wave lengths through the Whiteness Hp and Whiteness Hp Maxx gels.

The three factors studied in the experiment: gel, wavelength and time have a statically significant influence transmitting light into the gel.

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Influence of Boroxide Bioactive Bioglasses (BBB) on osteoblast viability

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ABSTRACT: The term bioglass addresses to surface reactive glass biomaterials. Their high bioactivity results in formation of strong bonds with the neighboring bones while the material contacts the bone. Bioactivity performance of bioglasses has been experimentally and clinically demonstrated both *in vitro* and *in vivo*, respectively. The first bioglass, named as bioglass 45S5, was invented by Hench with typical formulation of 25.5% Na₂O, 24.5% CaO, 45% SiO₂ and 6% P₂O₅. That classical formula melts, however, at around 1450°C. Addition of various amounts of B₂O₃ causes considerable lowering of glass melting temperature down to 950–1050°C. The significant reduction of glass melting temperature increases the feasibility of glass production processing and reduces the production cost of the resultant glasses. Under the perspective of health care, the primary use of boron in the body aims at absorbing calcium. Hence, boron is essential for healthy bones. In fact, boron is useful for women suffering from postmenopausal osteoporosis. Our study addresses its interest in boron-containing bioglasses, termed as boroxide containing bioactive bioglasses (BBB). In particular, the investigated glasses were based on the 45S5 bioglass having additives of B₂O₃ as follows: (a) Glass-B2 31.5% B₂O₃, (b) glass-B3 12.48% B₂O₃, (c) glass-B4 6.5% B₂O₃, (d) glass-B5 3.5% B₂O₃, and (e) glass-B6 2.5% B₂O₃. Medium, containing fine powders from each BBB glass, was put in contact with osteoblasts that have been plated at 1×10^3 cell density. The experiments were performed 72 hours after incubation. Morphological changes were investigated under light microscope. Cells' viability was assayed by MTT. Among the 5 investigated BBB glasses, the glass-B4 and glass-B5 exhibited the best performance and hence they can be qualified for further consideration and experimentation.

1 INTRODUCTION

Bioglasses, are highly bioactive ceramics and their bioactivity has been demonstrated both *in vitro* and *in vivo*, with the formation of a strong

bonding with the neighboring bone. However, medical applications of bioglass have been centered on low stress fields, mainly due to its non-adequate fracture toughness compared to that of cortical bone. This limitation is unfortunately

a common characteristic of glasses, ceramics and glass ceramics used in medical applications (Amaral *et al.*, 2002). The density of bioglasses is approximately 2.45 g/cm³; they have a hardness of 458 HV, mechanical strength of 100–200 MPa and 1.2–2.6 MPa m^{1/2} fracture toughness. They can carry on their mechanical properties in vitro and in vivo more space of time than HA (Keskin, 2002). Bioactivity of bioglasses is also higher than that of HA. It is assumed that the presence of bioactive glassy phase can provide faster response in terms of bone healing and bonding processes. It is also observed that glass-reinforced HA composites exhibit greater biological activities than commercial HA (Goller and Oktar, 2002). Bioglasses are highly bioactive, resulting in the formation of strong bonds with the neighboring bones. Their bioactivity has been demonstrated both in vitro and in vivo. Nevertheless, medical applications of bioglasses have been focused on cases of low stress fields, mainly due to their non-suitable fracture toughness, compared to that of cortical bone. Although both HA and bioglasses feature poor mechanical properties, limiting their applications in biomedicine (Goller *et al.*, 2003), it is expected that the mechanical properties can be considerably improved with additions of glass-ceramics introduced via sintering aids (Oktar and Goller, 2002) and incorporated into strong HA composition (Oktar *et al.*, 2007). Another study describes the production of some B₂O₃ containing bioglasses, which were melted between 950 and 1050°C (these low temperatures also aim at the economic production of such glasses) (Bogazici University Research Project). There are three bioglass preparation patents describing materials containing upto 5 wt% B₂O₃ [8–10] (Hench and Buscemi, 1978, Hench and Buscemi, 1979, Hench and Buscemi, 1980). In this study, the investigated glasses were (a) Glass B2-31.5% B₂O₃, (b) Glass B3-12.48% B₂O₃, (c) Glass B4-6.5% B₂O₃, (d) Glass B5-3.5% B₂O₃, and (e) Glass B6-2.5% B₂O₃. The BHA composites contained 5 wt% and 10 wt% Glass B5. Osteoblasts are cells that support the synthesis, secretion and mineralization of extracellular bone matrix. Therefore, the investigation of their behaviour in the presence of bioglasses is important to evaluate the biocompatibility of the tested glasses.

The purpose of this study is to find the best bioglass composition for grafting purposes.

2 MATERIALS AND METHODS

Bioglasses which were produced in a former study were used in this study (Bogazici University Research Project). Briefly, Analytical degree reagents were obtained from Merck Inc. SiO₂, CaO,

Table 1. Composition of each sample.

1st composition (melted at 1000°C)			
SiO ₂	40 gr.	40%	SiO ₂
CaO	14 gr.	14%	CaO
Na ₂ B ₄ O ₇	45.5 gr.	14%	Na ₂ O
		31.5%	B ₂ O ₃
TiO ₂	0.5 gr.	0.5%	TiO ₂
	100 gr.	100%	
2nd composition (melted at 950°C)			
SiO ₂	50 gr.	50%	SiO ₂
CaO	14 gr.	14%	CaO
Na ₂ B ₄ O ₇	30 gr.	23.07%	Na ₂ O
NaCO ₃	18 gr.	12.48%	B ₂ O ₃
TiO ₂	0.45 gr.	0.45%	TiO ₂
	112.45 gr.	100%	
3rd composition (melted at 1050°C)			
SiO ₂	50 gr.	49.6%	SiO ₂
CaO	6.96 gr.	14%	CaO
Na ₂ B ₄ O ₇	9.4 gr.	23.6%	Na ₂ O
NaCO ₃	34.2 gr.	6.5%	B ₂ O ₃
Ca(PO ₄) ₂	13.1 gr.	6%	P ₂ O ₅
TiO ₂	0.5 gr.	0.5%	TiO ₂
	114.16 gr.	100.2%	
4th composition (melted at 1000°C)			
SiO ₂	52 gr.	52%	SiO ₂
CaO	6.9 gr.	14%	CaO
Na ₂ B ₄ O ₇	5.06 gr.	23%	Na ₂ O
NaCO ₃	36.67 gr.	3.5%	B ₂ O ₃
Ca(PO ₄) ₂	13.1 gr.	6%	P ₂ O ₅
TiO ₂	0.75 gr.	0.75%	TiO ₂
CaF ₂	0.75 gr.	0.75%	CaF ₂
	115.23 gr.	100%	
5th composition (melted at 1000°C)			
SiO ₂	52.5 gr.	52.5%	SiO ₂
CaO	6.9 gr.	14%	CaO
Na ₂ B ₄ O ₇	3.62 gr.	23%	Na ₂ O
NaCO ₃	37.42 gr.	2.5%	B ₂ O ₃
Ca(PO ₄) ₂	13.1 gr.	6%	P ₂ O ₅
TiO ₂	1 gr.	1%	TiO ₂
CaF ₂	1 gr.	1%	CaF ₂
	115.54 gr.	100%	

Na₂B₄O₇, NaCO₃, Ca(PO₄)₂, TiO₂, CaF₂ reagents were mixed at certain percentages at a 55 gr. Pt crucible as described on Table 1. The mixtures were well homogenized in a mortar separately and then melted in a Pt-crucible at 950–1000°C for 1 hour. The molten glass compositions were quenched by pouring into cold water. The glass-frits were properly ball-milled separately to obtain fine powder mixtures.

2.1 Culture of osteoblasts

Penicillin, streptomycin, fetal bovine serum: Gibco (NY, USA), Dulbecco's phosphate buffered saline, trypsin-EDTA, [3(4,5 dimethylthiazol-2yl) 2,5 diphenyltetrazoliumbromide] MTT, BCIP-NBT kit: Gibco (NY, USA). Crude bacterial

collagenase: Boehringer (Biberach, Germany). RPMI Cell culture medium: Sigma (St Louis, USA), SIRCOL kit: Biocolor (Newtonabbey, N Ireland) T25 culture flasks and multidish 24 wells: Nunc products (Naperville, USA).

Osteoblasts were isolated from the calvaria of 1–5 days old neonatal Wistar rats. The calvaria was dissected and freed from soft tissue, cut into small pieces and rinsed in sterile phosphatebuffered saline without calcium and magnesium. The calvaria pieces were incubated with 1% trypsin-EDTA for 5 minutes, followed by four sequential incubations with 0.2% collagenase at 37°C for 20 minutes each. The supernatant of the first collagenase incubation, which contain a high proportion of periosteal fibroblasts, was discarded. The other digestions produced a suspension of cells with high proportion of preosteoblasts and osteoblasts. After centrifugation at 1400 g for 5 min, each pellet was resuspended in 5 ml of RPMI medium supplemented with 10% FBS, 1% antibiotic-antimycotic. The cells were seeded into 25 ml tissue culture flasks, and lead to grow in a controlled 5% CO₂ 95% humidified incubator at 37°C. After confluence the cells were used for experiments on passage 2.

2.2 Cellular viability

Osteoblasts from the second passage were plated on 24 wells culture dish at a 1×10^4 cell density. After seeding the medium was changed for medium containing the glasses powders and the control cells had the medium changed for fresh one. After 72 hours of incubation osteoblast viability was evaluated by MTT assay, based on the reduction of tetrazolium salt to formazan crystals by dehydrogenase present in living cells mitochondria. Sixty μ l of MTT (5 mg/ml) was added to each well. Two hours later, the cell morphology was analyzed by inverted optical microscopy and formazan salts were solubilized with 100 μ l SDS 10% HCl. After incubation for 18 hours the optical density measurement was done at 595 nm. The results were plotted as control percentage.

2.3 Alkaline phosphatase production

The alkaline phosphatase production was evaluated by BCIP-NBT assay which is based on a chromagenic reaction initiated by the cleavage of the phosphate group of BCIP by alkaline phosphatase present in the supernatant. This reaction produces a proton, which reduces NBT to an insoluble purple precipitate. Briefly, the cells were incubated in the presence of the glasses following the same protocol described for viability test. After 72 hours, the supernatant of each well was removed and 100 μ L

of PBS was added to each well. Then, 50 μ L of BCIP-NBT solution was added to each well according to the manufacturer's protocol. After overnight incubation at 37°C, the formed blue crystals were solubilized and the optical density measurement was done at 595 nm. The reserved supernatant was also evaluated by adding 50 μ L of NBT-BCIP solution to 100 μ L of the supernatant. After overnight incubation the absorbance were measured at 595 nm. The results are presented as means $\pm$ SD. The statistical significance was measured by ANOVA and Bonferroni's post-test (confidence level was 95%).

3 RESULTS AND DISCUSSION

Considering that osteoblasts are the cells that support the synthesis, secretion and mineralization of extracellular bone matrix, the investigation of their behavior in the presence of bioceramics is important to evaluate biocompatibility. The bioceramics composition, cristallinity, and porosity are characteristics directly related to cell physiology (Amaral *et al.*, 2002) and the ionic products from bioceramics dissolution are responsible for alterations in osteoblast proliferation (Keskin *et al.*, 2000). So, in this work we investigate the viability and proliferation of osteoblasts in the presence of different groups of BorGlass.

Under light microscopy, there were no osteoblasts morphological changes observed each 24 hours, during 3 days. The viability and proliferation of osteoblasts had no significant difference in the presence of group I, II, and III, when compared to control. Viability and proliferation was increased in the presence of group VI and V when compared to control and to the other groups.

The presence of Borglasses enhanced viability and proliferation of osteoblasts when compared to control in two formulations up to 50% (Fig. 1).

Of few studies about the contribution of boron to biomaterials, some are very negative and some of seem quite promising. These results and discussion must do for the future studies. The primary use of boron in the body is for the efficient absorption of calcium, and thus, boron is essential for healthy bones. Boron is useful for women suffering from postmenopausal osteoporosis. Boron is essential for the utilization of vitamin D, which enhances the absorption of calcium. Recent research demonstrates that boron may be essential in the conversion of vitamin D to its active form (<http://www.oralchelation.com/ingred/boron.htm>). Sheng *et al.* have demonstrated that boron as dietary boron supplementation affects the bone mineral balance in rats (Sheng *et al.*, 2001).

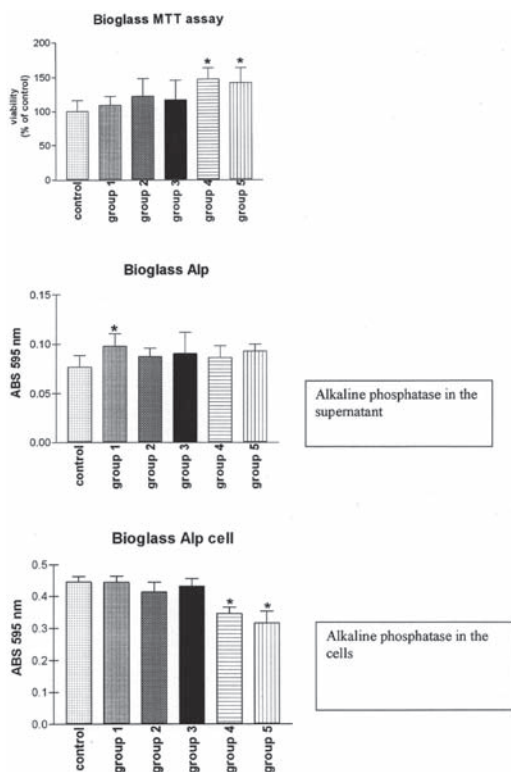


Figure 1. a) Results of cell viability tests: Osteoblasts were plated in the presence of granules of the five different investigated BorGlasses (Group 1 = B2, Group 2 = B3, Group 3 = B4, Group 4 = B5, Group 5 = B6). After 72 hours of incubation, viability was evaluated by MTT assay and compared to control. Osteoblasts showed a significant increase on viability/proliferation in the presence of the glasses B5 (group 4) and B6 (group 5), when compared to control. Results represent mean $\pm$ SD of 3 triplicates from 3 separate experiments ($P < 0.05$). b, c) Results of alkaline phosphatase production: Analyzing the amount of alkaline phosphatase in the cells and in the supernatant it was demonstrated that the presence of the glasses delayed the alp production by osteoblasts.

Dupre *et al* have also demonstrated that of boron additions to the diet of vitamin D-deficient chicks indicated that boron plays an important role in animal nutrition (Dupre *et al.*, 1994). McGrath has shown that boron can be used as a suitable material for coating orthopedic implants (http://materialsknowledge.org/index.php?option=com.docman&task=doc_download&gid=23&mode=view). Kitsugi *et al.* added boron to some very popular glass ceramic compositions and observed that boron neither promoted the dissolution of the glass-ceramics nor influence the bone formation

at the interface of the ceramic and bone (Kitsugi *et al.*, 1992). Vrouwenvelder *et al.* has used a cell culture model to compare Bioglass 45S5 with four other bioactive glasses. Small substitutions or additions of certain ions like iron, titanium, fluorine or boron modified the basic 45S5 glass network (Vrouwenvelder *et al.*, 1994). Silva *et al.* has used boron carbide as a substitution material in biofoams. The Bioglass and bioglass+B4C shows a lower density, in foam-glass form and has the bioactivity characteristic in a SBF and blood suspensions in-vitro assays. They have reported that this new composition could be used for biomedical applications (<http://www.shimadzu.com.br/analitica/aplicacoes/difratometros/raios-x/xrd/bioglass-2.pdf>). Our findings of no alteration in cell morphology are in accord to the literature that describes the bioactive glass as the most biocompatible biomaterial (Goller *et al.*, 2003). Our results showed an interesting performance of sample 4 and 5 enhancing osteoblast viability. It is important to point that the other samples did not enhance but also did not impair the viability of the osteoblasts indicating that all formulations could be considered for further investigations depending on the purpose of the experiment. If you want cell proliferation, use sample 4 or 5. If you want just to maintain the control proliferation rate, use the other samples. On the other hand we can state that the sample 4 and 5 delayed the differentiation of the osteoblasts since the alkaline phosphatase production was decreased in the presence of these samples. Alkaline phosphatase production is an important marker of early osteoblast differentiation (Bruder *et al.*, 1998). If we consider that the number of viable cells was much higher in the presence of samples 4 and 5, the alkaline phosphatase production was significantly impaired. This finding also can be used for healing purpose, considering that in some cases the mineralization delay would allow better matrix regeneration.

4 CONCLUSION

Since it is known that the silicon contents (Langer, 2000; Valerio *et al.*, 2004) and the sintering temperature (Valerio *et al.*, 2004) can interfere with biocompatibility, we speculate that the observed increase in proliferation is due to a proper silicon content associated with an adequate sintering temperature. Our results indicate the Borglass for further investigations for use in bioengineering

Our results strongly suggested that the two tested bioactive glasses compositions have the property of increasing the osteoblast proliferation, since MTT assay gives an indirect measurement of cell proliferation based on cell viability.

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Effect of sintering temperature on mechanical properties and microstructure of zeolite (clinoptilolite) reinforced bovine hydroxyapatite (BHA) composites

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ABSTRACT: The applications of ceramics of pure hydroxyapatite (HA), which are nowadays restricted to non-load bearing implants due to the poor mechanical properties of HA, can be expanded via doping of HA-matrix with biocompatible (or even better bioactive) oxides resulting in strong HA-composites. In this study, we have doped biologically derived apatite from bovine bones (BHA) with 5 wt% and 10 wt% zeolite, clinoptilolite, which is a very pure form of zeolite, and being currently widely used in drug manufacturing industry, in poultry (for getting eggs with better properties) and feeding ingredient for cows for getting high quality milk. The HA-zeolite composites were produced via regular sintering technique, whereby clinoptilolite was used considering to serve as a wetting agent of HA particles. The compression strength and the microhardness of produced composites were measured as 127.7 MPa for 10% zeolite addition and sintering at 1200°C, and 431 HV for 10% zeolite addition and sintering at 1300°C.

1 INTRODUCTION

Musculoskeletal system diseases cost countries around the world a big amount of money. Moreover, the occurrence of bone fractures has increased due to an increasing number of traffic accidents as well as increase of life expectancy from age of 35 to 70 (Ozyegin *et al.*, 2004). For example, US National Highway Traffic Safety Administration (2003) has reported that the use of seat belt saved more than 100,000 lives, but over 7,000 people were killed and over 100,000 people

were injured because of not using seat belt in only in the U.S. The study of Simsekoglu and Lajunan showed that in Turkey in 2005, 570,419 traffic accidents happened, in which 3,215 people died and 123,985 people got injured. The worsening of these numbers over the time clearly indicates the need for producing new efficient and low-cost biomaterials (Simsekoglu and Lajunen, 2008). On the other hand, human life will increase to 90 years in the next 25 years. There are many reports claiming that in Europe, the number of adults will increase by 50% during the next 50 years where one third of

European citizens will be over the age of 60. With this greater life expectancy, healthcare costs will be an increasing burden for society. The need for biomaterials, which will be economic, simple and feasible, is certainly a demanded challenge for the beginning of the 21st century.

Under that general perspective, bioceramic materials are very important. Bioceramics are widely used to repair and reconstruct damaged parts of human skeleton especially as bone substitutes at filling of bone defects. Calcium phosphate ceramics and in particular hydroxyapatite (HA) have received great attention in clinical practice. Because of their excellent biocompatibility, bioactivity and osteoconduction properties, these compounds are increasingly used as bone substitute materials (Descamps *et al.*, 2009). However, the applications of ceramics made of pure HA are limited to non-load-bearing implants, such as inner-ear bones or filler materials because of their poor mechanical properties (Erkmen *et al.*, 2007). Doping with biocompatible or even better bioactive oxides may result in strong HA-composite materials (Oktar *et al.*, 2007). Hence, there is special interest for reinforcing a matrix of apatite with other materials, such as by incorporating ceramic or metallic fibers, whiskers, platelets or particles, to expand the use of the resultant HA-composites in load-bearing implants. Thin-coatings of HA on metal substrates (e.g. by plasma-spray) is a special form of such composites (Gunduz *et al.*, 2009).

There are very few papers reporting the use of zeolites as biomaterial (Ceyhan *et al.*, 2007; Schainberg *et al.*, 2005). The use of zeolites in medicine is a relatively recent subject of interest that is emerging at a sound pace. In particular, the natural zeolite clinoptilolite has been the main subject of interest for such applications due to their abundance and the variety of ions in the structure. The inclusion of zeolite into animal diets has been proven to be useful against the toxic effects of aflatoxins and mycotoxins. Zeolites may be used as antibacterial agents when especially Ag is also incorporated into these materials by ion-exchange. The encapsulation of different ions and molecules in zeolites and then the slow release of these materials provide the opportunity of zeolites to be used in drug complexation. Zeolites may also be utilized as biosensors (Ceyhan *et al.*, 2007). Some researchers had used also zeolite as a graft material instead of HA material (Schainberg *et al.*, 2005) and other biomedical applications like skin treatment, diarrhea treatment and many others (Ozyegin *et al.*, 2003).

In this study, we have doped biologically derived HA from bovine bones (BHA) with (5 and 10 wt%) clinoptilolite.

2 MATERIALS AND METHODS

Bovine bones (femoral parts) were calcinated at 850°C for 4 h in air as reported in earlier studies (Oktar *et al.*, 1999; Goller *et al.*, 2006). Calcination at high temperature was done to totally eliminate any risk of transmitting diseases (Ozyegin *et al.*, 2004; Goller *et al.*, 2004). Calcinated bone-pieces were crushed and then ball-milled until fine powder (i.e. with particles of submicron average size) of apatite (BHA) was obtained. The resultant fine powder of BHA was admixed with 5 wt% and 10 wt% fine clinoptilolite.

The prepared mixtures were pressed to pellets with diameter and height of 11 and 12 mm, respectively, by uni-axial cold pressing, according to the British Standards and then sintered at different temperatures between 1000°C and 1300°C for 4 h in air. Heating and cooling rates were 2 K/min. Measurements of compressive strength (σ , Instron 8511, displacement 2 mm/min), Vickers microhardness (measured in HV units in a Shimatzu apparatus with 200 g load), and density (d) by Archimedes immersion method (equipment Precisa), along with observations of the microstructure by scanning electron microscopy (SEM, JSM-5410) were performed on the sintered samples. X-ray diffraction studies were also performed (Bruker).

3 RESULTS AND DISCUSSION

The high temperature of calcination during BHA production (850°C) implies that any possible disease existing in the original bone has been completely eliminated. It has been mentioned that at high temperatures no prion diseases can survive. Earlier reports have suggested that interesting bioceramics can be produced from high-temperature calcined BHA; this has also been confirmed by cell culture and animal studies.

For getting more loads resistable HA ceramics, usually a secondary phase must be added. The results of the present study, summarized in Table 1, being compared with results of earlier

Table 1. Influence of zeolite content and sintering temperature on compression strength (σ) and Vickers microhardness of BHA-zeolite composites.

T (°C)	5%		10%	
	σ (MPa)	HV	σ (MPa)	HV
1000	30.6	88.3	50.1	105.5
1100	44.9	90.8	68.02	76.2
1200	70.5	146	127.7	211
1300	136.6	247	20.73	431

studies, reported in Tables 2 and 3, support this statement. The highest compression tests values were obtained after sintering at 1300°C using 10% clinoptilolite addition (127.7 MPa) and 5% addition (136.6 MPa). Goller *et al.* obtained 67 MPa [9] and Goren *et al.* & Gokbayrak 80 MPa at 1200°C (Table 2) (Goren *et al.*, 2004; Gokbayrak 1996).

Since clinoptilolite is an aluminum-silicate compound, it has the ability of wetting effect of the particles of HA during sintering process at high temperatures. This is clear by looking at the SEM-images of Figures 1–4. The images corresponding to sintering at 1300°C clearly show the big crystals developed after sintering. The effect of good wetting is seen whereby the particles are

Table 2. Compression strength (σ) & microhardness of BHA-CIG (commercial inert glass) composites reported in earlier studies.

T (°C)	5% CIG		10% CIG	
	σ (MPa)	HV	σ (MPa)	HV
1000	48.32	98.3	64.02	53.92
1100	43.89	149.2	77.04	134.93
1200	67.58	202.3	132.98	506.5
1300	104.81	227.3	6.86	306.63

Table 3. Compression strength (σ) and microhardness results reported in earlier studies for samples of pure BHA ceramics sintered at different temperatures.

T (°C)	BHA [9]		BHA [11, 12]	
	σ (MPa)	HV	σ (MPa)	HV
1000	12	42	52	85
1100	23	92	22	70
1200	67	138	80	150
1300	60	145	70	130

well rounded with broad borders of glassy phase. In the literature there are many studies reporting addition of glass compositions for getting a better densification and wetting effect in HA-glass composites. Oguzhan *et al.* and Salman *et al.* had done so. Their best results (Table 2) were

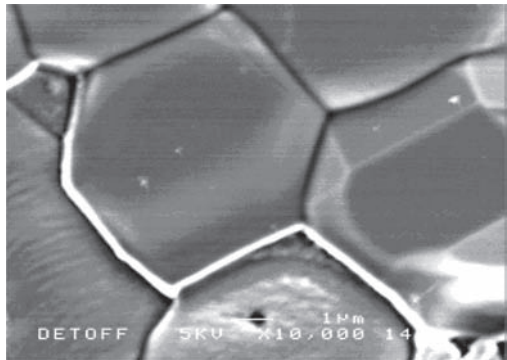


Figure 2. BHA-5 wt% zeolite sintered at 1300°C.

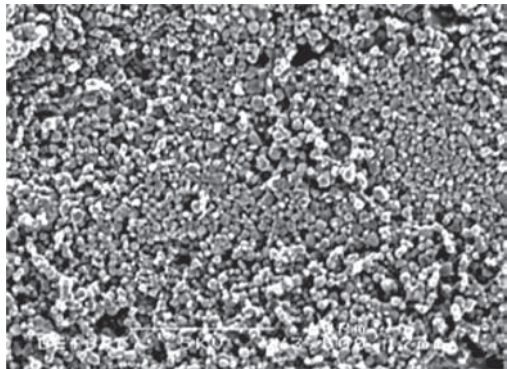


Figure 3. BHA-10 wt% zeolite sintered at 1000°C.

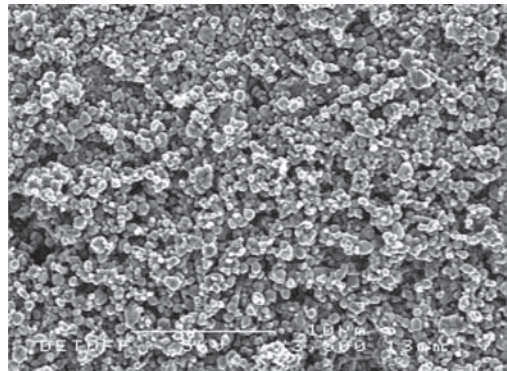


Figure 1. BHA-5 wt% zeolite sintered at 1000°C.

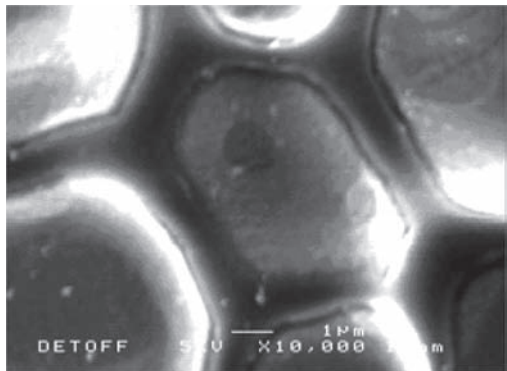


Figure 4. BHA-10 wt% zeolite sintered at 1300°C.

138.98 MPa with 10% glass-addition (sintering at 1200°C) and 104.81 MPa with 5% glass-addition (sintering at 1300°C) (Gunduz *et al.*, 2009; Salman, 2007). X-ray diffraction studies were indicating major HA phases with some minor HA alumina silicate phases.

As far as the nature and general features of zeolites are concerned, they are structurally hydrated microporous crystalline aluminosilicates that can occur naturally or being synthesized in laboratory conditions. Zeolite framework consists of an assemblage of SiO_4 and AlO_4 tetrahedra, joined together in various regular arrangements through shared oxygen atoms, to form an open crystal lattice. Zeolite's 90% of the active compound consists of clinoptilolite (Aluwéa *et al.*, 2009). With regards to biological performance, Keeting *et al.* have concluded that zeolite induces the proliferation and differentiation of cells of the osteoblast lineage (Keeting *et al.*, 1992).

4 CONCLUSIONS

Clinoptilolite seems to be a very promising reinforcement material for adding in HA compounds. Clinoptilolite material improves wetting behavior and mechanical properties. The best compression strength values were 127.7 MPa achieved for 10% addition and sintering at 1200°C and 136.6 MPa achieved for 5% addition and sintering at 1200°C. The best microhardness values were 247 HV after sintering at 1200°C of the samples with 5% addition and 432 HV after sintering at 1300°C of the samples with 10% addition indicating that the higher amount of zeolite, the higher the microhardness.

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The thickness of the cortical bone in different maxillae using medical images

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ABSTRACT: The aim of this study was to investigate whether there is a relationship between the thickness of the cortical bone of mandible human and the age and the sex of patient. In this work the measure of the cortical bone thickness was obtained in different computed tomography (CT). Different human mandibles were scanned using high resolution micro-CT instrument in which many axial slices were obtained. A total of four medical images were studied and observed. Two different groups were characterized. The first one, with two female maxillae (F): an old and a young patient. The second group of two males mandibles (M), with similar age. A comparison between the male and female sex was also obtained. The cortical bone thickness of the mandible may be affected by tooth extraction, age and sex patient. The use of this type of information is useful for complementary diagnostic information and treatment planning.

1 INTRODUCTION

CT is the most common technique used for examination of maxillofacial because it permits the visualization of soft tissues and bone structures in the same medical image (Cavalcanti et al. 2001). This technique is used in several clinical dentistry applications even by axial slices and two (2D) and three-dimensional (3D) reconstructed images (Rocha et al. 2003).

In some regions of the mandible is very difficult to distinguish between cortical and trabecular bone (Natali et al. 2003). Therefore, with thresholding, segmentation operations and to built separate models, the modelling procedure is possible using dedicated software package. Many researchers have studied the relation between CT values and the bone material properties.

The cortical bone at the alveolar bone ridge is in general much thinner than the basal bone (Natali et al. 2003), and generally much lower than CT values of cortical bone at other locations.

Tooth extraction causes continuous and irreversible bone reduction at the mandible (Polat

et al. 2001). A number of factors affect the bone and cortical thickness, hormonal, metabolic, endocrinology and dietary factors.

With this work many conclusions will be produced through the analysis of four different human mandibles.

2 METHODOLOGY

Different layers from each human mandible were selected to measure the number of pixels values. Pixel by pixel for each slice and using image control system software, different layers were measured.

A total of four medical images were studied and observed. The measure was made since bottom mandible until above.

A CT high resolution was used and the pixel value is equal to 0.269531 mm. Four different CT were analysed.

Figure 1 represents different axial planes scans of each female maxilla in study, considering the outer and inner side for the points of measurement.

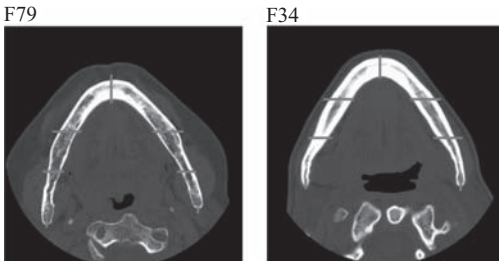


Figure 1. 2D scans of female maxillae and points for measure the cortical bone (F79 and F34).

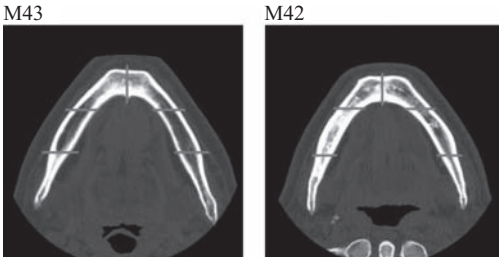


Figure 2. 2D scans of male mandibles and points for measure the cortical bone (M43 and M42).

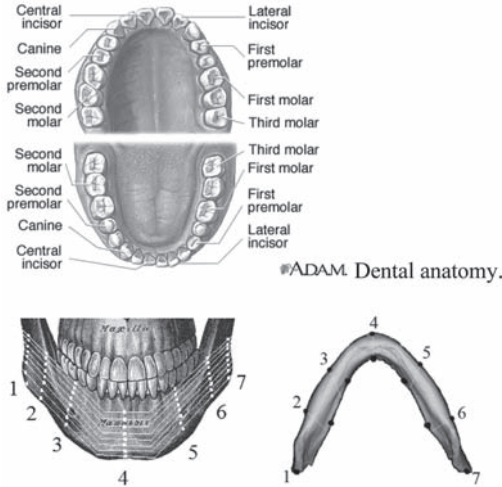


Figure 3. Dental anatomy. Layers and seven measured points. (Courtesy of ADAM).

The axial planes scans of male mandibles are represented in figure 2. The points of measurement are also considered for outer and inner side.

The numbers of teeth are similar between male and young female mandible. Only the old female maxilla has no teeth.

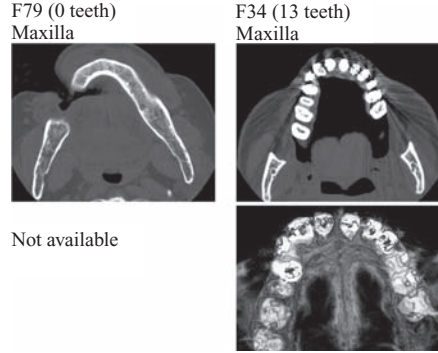


Figure 4. Number of teeth, 2D and the 3D scan of female maxilla for each study case.

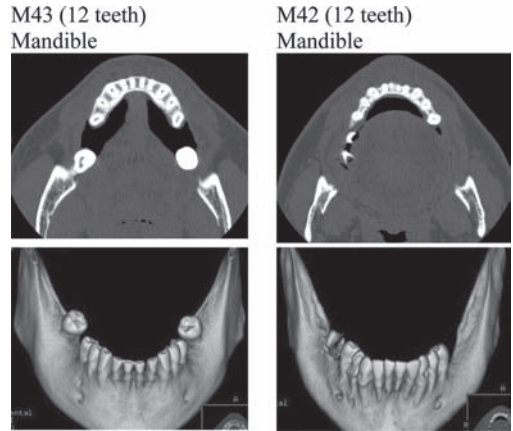


Figure 5. Number of teeth, 2D and the 3D scan of male mandible for each study case.

For female group maxillae were studied and inferior maxillae or mandibles for male group. The patient age is given close by each medical image.

The cortical bone thickness was calculated from the outer to the inner side in seven different portions for different layers, through the measure of pixels numbers.

Figure 3 shows a dental anatomy and different selected seven points along each mandible and maxilla.

All studied cases are represented in figures 4 and 5. A 3D scan of each case was represented. The CT visualization was made with GE medical system DICOM, a viewer Software.

3 RESULTS

In this work 2 females maxillae and 2 males mandibles CT were studied.

Table 1. Mean values of cortical thickness in different mandibles for outer side, mm.

Measured points	F79 (0 teeth)	F34 (13 teeth)	M43 (12 teeth)	M42 (12 teeth)
1	0.977	2.021	1.168	2.366
2	1.715	2.734	3.264	3.658
3	1.348	1.920	2.036	1.685
4	1.579	2.387	1.201	1.583
5	1.572	1.977	1.563	1.482
6	1.572	2.464	2.931	2.426
7	0.931	2.313	0.988	1.707
CTM	1.39	2.26	1.88	2.13

Table 2. Mean values of cortical thickness in different mandibles for inner side, mm.

Measured points	F79 (0 teeth)	F34 (13 teeth)	M43 (12 teeth)	M42 (12 teeth)
1	0.977	2.021	1.168	2.366
2	0.515	1.386	1.737	1.887
3	1.348	1.887	2.336	2.325
4	3.157	2.734	2.867	2.628
5	1.168	1.977	1.941	1.853
6	0.786	1.040	1.314	1.437
7	0.931	2.313	0.988	1.707
CTM	1.27	1.491	1.76	2.03

Different axial slices were selected, as can see in figure 3, and seven measurements were made for each other. The estimated value was the mean of these calculations.

The average values of cortical thickness are given in the table 1 and 2, according the age and sex patient for each measured point.

A cortical thickness of maxillae (CTM) is also represented in tables 1 and 2.

CTM is the mean of all collected values from the measured points for one maxilla.

The values of cortical thickness are calculated for outer and inner side of mandibles, as represented in figures 1 and 2.

4 DISCUSSION AND CONCLUSIONS

The CTM measurements in tables 1 and 2 show significant differences between the age and the tooth extraction.

As can see in figure 6, F79 patient has the lesser value. CTM greater value is for F34 patient. The CTM values for M patients are similar and approximately equal 2 mm.

Figure 7 represents the values for inner side in F group. The value with more difference between inner and outer side is the obtained for F34 patient.

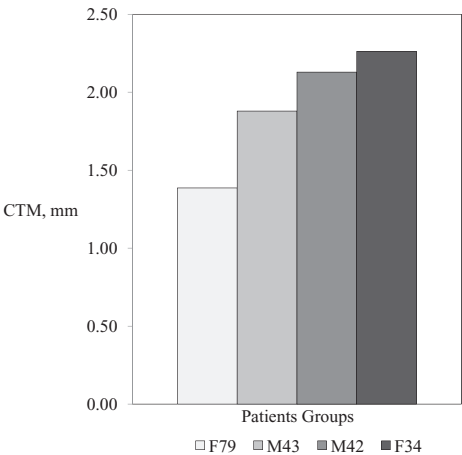


Figure 6. CTM values for F and M patients, (outer side).

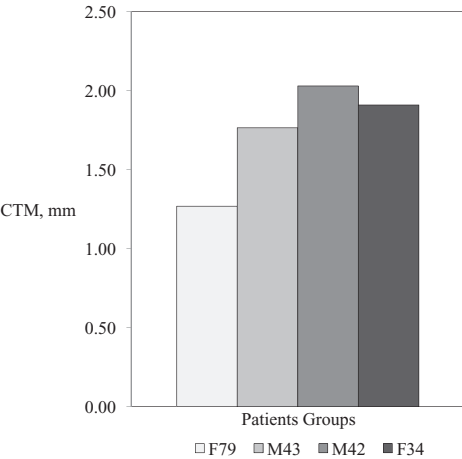


Figure 7. CTM values for F and M patients, (inner side).

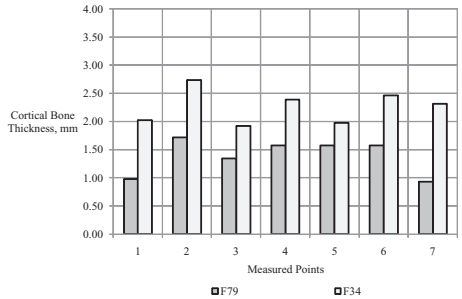


Figure 8. Values of cortical bone in F group, (outer side).

Figures 8 and 9 represent one graphic with the comparison of cortical bone thickness between different ages for F group, for outer and inner side of maxillae, respectively.

Figures 10 and 11 represent the comparison of cortical bone thickness in similar ages for M group.

As can see, the cortical thickness for old F patient is lesser than young F patient. The number of teeth in the F34 maxilla was also higher when compared with F79. The teeth extraction results in a decrease in the resistance of the maxillae. The values of cortical thickness for F maxillae have a constant behaviour for outer side measure. The range of F79 cortical thickness is between 1–1.6 mm. For F34 the values are in the 2 and 2.75 mm.

The values for M groups are more similar, except for the lateral points (1 and 7). The young

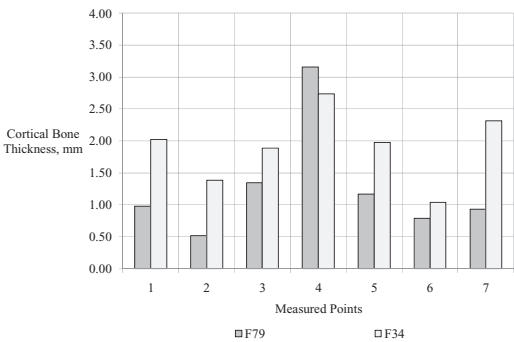


Figure 9. Values of cortical bone in F group, (inner side).

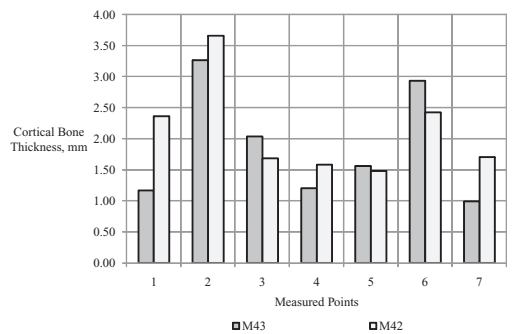


Figure 10. Values of cortical bone in M group, (outer side).

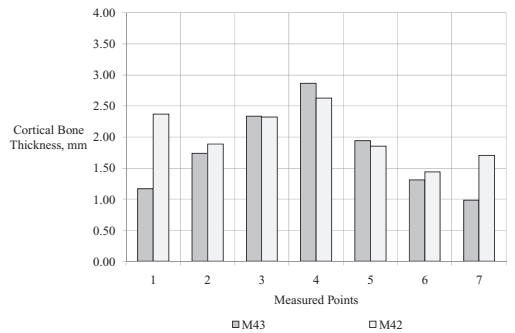


Figure 11. Values of cortical bone in M group, (inner side).

male and female maxillae present a relative differences and behaviour. For M group the range values have a greater oscillation and are between 1 and 3.75 mm.

In the middle of measure, and independently of sex and age, the cortical bone thickness is greater in inner side of mandibles.

More CT analysis should be analysed to determine the correlation between different groups and sex. The cortical thickness can be calculated in conventional radiographs, but the use of CT method is an excellent way to distinguish different tissues, in this case, to distinguish the inner and outer cortex.

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Glass-ionomer cements: A review of their engineering properties

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ABSTRACT: Glass-ionomers are shown to have acceptable mechanical properties, as well as bonding directly to dentine and enamel. Modern cements have been subject to improvements, particularly in particle size and particle size distribution. They can therefore be used in more demanding applications such as the Atraumatic Restorative Treatment (ART) technique and Minimal Intervention (MI) dentistry, and also in load-bearing parts of the mouth. Adhesion and durability of these materials are generally good.

1 INTRODUCTION

1.1 Clinical applications

Glass-ionomer cements have been used for several years now in a variety of clinical applications within dentistry (Mount 2002). These include as liners and bases, and as direct filling materials. There are also brands for use as bonding agents in orthodontics and as sealants in endodontics.

1.2 Composition and setting

Glass-ionomers consist of powdered ion-leachable glasses reacted with aqueous acidic polymer solutions, either 45% polyacrylic acid or an equivalent concentration of acrylic acid/itaconic acid copolymer (Wilson & McLean 1988). In addition, (+)-tartaric may be present to control the rate of setting (Wilson et al. 1976). Reaction occurs quickly (in circa 5 minutes or less) and the cement sets to a rigid material with a slightly translucent appearance.

Typical glass compositions are shown in Table 1. Glasses are basic, as a result of aluminium being forced to adopt 4-co-ordination by the excess of silica in the structure. NMR studies have

demonstrated that glasses have significant quantities of 4-co-ordinate aluminium (as shown by a peak at 54–56 ppm), but that this changes to give predominately 6-co-ordinate aluminium (peak at –2 ppm) in the set cement (Zainuddin et al. 2009).

Setting also involves formation of calcium and aluminium polyacrylate units (Crisp et al. 1974), and there seems to be a role for the ion-depleted glasses, which may result in the development of an additional inorganic network in the set cement (Wasson & Nicholson 1993).

Freshly set cements are sensitive to water loss, which may cause surface micro-crazing and loss of aesthetics. To prevent this, in clinical use, glass-ionomers are protected with a coat of varnish or petroleum jelly. This reduces net water loss, and also alters the mechanism by which water escapes (Nicholson & Czarnecka 2007), with the net effect that the surface remains undamaged and the appearance good.

After the initial hardening, maturation reactions take place gradually. These lead to improvements in translucency and compressive strength, and a reduction in the amount of labile (“unbound”) water within the cement (Nicholson 1998).

2 ENGINEERING PROPERTIES

2.1 Strength

The current ISO standard requires glass-ionomers to have a minimum compressive strength of 70 MPa for application as liners or bases, and 150 MPa for use as a direct restorative material (ISO 2007). Modern brands are easily able to exceed these minima, with values in the range 220–300 MPa being commonly achieved (Guggenberger, May & Stefan 1998).

However, the materials are brittle and the use of compressive strength as the criterion of

Table 1. Typical compositions of glasses for glass-ionomer cements (%).

Element	G200	G338	MP4
Si	13.9	11.8	13.1
Al	13.4	16.9	18.5
Ca	17.6	6.6	18.6
F	20.1	19.7	–
Na	2.0	6.3	8.2
O	30.6	32.5	41.6
P	2.5	6.2	–

Table 2. Fracture toughness values (MN/m^{3/2}) for glass-ionomer cement (Ryan, Orr & Mitchell 2001).

Fuji I stored in		
Air	100% Relative humidity	Under water
0.27 (0.04)	0.27 (0.01)	0.26 (0.08)

acceptable mechanical properties for clinical use is questionable (Ban & Anusavice 1990). More important in this regard is fracture toughness (Hill, Wilson & Warrens 1989). There are practical difficulties in determining fracture toughness for glass-ionomers, and few studies have been reported on this property. However, there are some published results (Ryan, Orr & Mitchell 2001) and examples are listed in Table 2. Results show little fluctuation with storage conditions, though values are lower than those obtained for resin-modified glass-ionomers.

Diametral tensile strength has also been determined on these materials, using cylindrical specimens loaded along their long axis (Cefaly et al. 2003). Typical results for conventional glass-ionomers were in the range 8–15 MPa, which were again significantly lower than those found for resin-modified glass-ionomers.

In all cases, mechanical properties are found to vary with powder:liquid ratio of the cement and molecular weight of the polymer employed.

2.2 Adhesion

Glass-ionomers are able to bond naturally to the tooth surface, both enamel and dentine. Initial bond strengths are low compared with bonded composite resins, typically 1.5–5.0 MPa (Walls, McCabe and Murray 1988), but durability is greater (Tyas & Burrow 2004). Longer term studies have shown that there is an ion-exchange process at the interface between the cement and the tooth, and that this results in the formation of a distinct interfacial region with time. The formation of this interfacial region, termed an ion-exchange layer, has been claimed to be the reason for the long-term durability of the adhesive bond formed by these cements (Ngo et al. 1997).

2.3 Other properties

Glass-ionomers release fluoride (Forsten 1998). This has been shown to continue for several years in *in vitro* conditions, though to be reduced in the presence of saliva (Bell et al. 1999).

Release has been modeled mathematically and, under neutral conditions, shown to follow the equation:

$$[F]_c = K + \beta t^{1/2} \quad (1)$$

where K and β are constants, $[F]_c$ is the amount of fluoride released at time t (De Moor, Verbeeck & De Maeyer 1996).

This diffusion-based process becomes modified under mildly acidic conditions, or following exposure to aqueous fluoride solution, and a slow dissolution-based mechanism takes over (Dhont, De Maeyer & Verbeeck 2001).

Glass-ionomers will also take up fluoride under appropriate conditions (Forsten 1998). This uptake has been shown to follow pseudo-first order kinetics, with rate constants varying between 2.11×10^{-5} and $7.98 \times 10^{-5} \text{ s}^{-1}$, depending on the cement and the initial concentration of the fluoride solution (Pawluk et al. 2008). This has led to the suggestion that glass-ionomers can act as fluoride reservoirs though the one experimental study of fluoride uptake showed that generally only a fraction of the fluoride taken up was released in 24 hours on storage in pure water (Pawluk et al. 2008).

Glass-ionomers are also known to buffer mouth acids, and to be capable of shifting the pH of films of lactic acid from 4.5 (active caries) to 5.5 (arrested caries) within 1 minute (Nicholson et al. 2000). This buffering is associated with release of other ions, *i.e.* Na^+ , Ca^{2+} and Al^{3+} , and also Si and P species, some of which are useful in remineralization of the tooth surface (Nicholson et al. 2002).

3 RECENT DEVELOPMENTS

3.1 Atraumatic Restorative Treatment (ART)

ART is a clinical technique that has been developed mainly for use in the 3rd World (Frencken et al. 1996). In these regions, electricity supply is absent or unreliable, hence conventional dental drills cannot be deployed. On the other hand, levels of dental caries are increasing. Under a World Health Organisation initiative ART has been developed to address these problems. It involves use of hand-operated spoon-shaped scoops, designed to remove tooth material softened by caries. Repair is then carried out by placing a glass-ionomer cement.

Clinical results show that this overall approach is being very successful (Frencken et al. 2004). It has also been used for special-needs patients to allow home treatment in developed countries (Czarnecka 2006).

3.2 Glass-ionomers for ART

Glass-ionomers have been improved for this application, mainly by altering particle size and particle size distribution, and this has improved both compressive strength and toughness. These materials appear to have improved durability (Covey 2000).

Table 3. Compressive strength of teeth at various stages of cavity repair (Bharadwaj, Solomon & Parameswaran 2002).

Teeth	Compressive strength (MPa)	(S.D.)
Sound teeth	104.65	(13.59)
Cavity prepared, unrestored	48.88	(6.75)
Cavity repaired with light-cured composite	73.62	(15.52)

3.3 Minimal Intervention (MI) dentistry

This is a multi-faceted approach to clinical dentistry based on a combination of early diagnosis of caries, enhanced remineralization, and repair with smaller cavity sizes (Mount & Ngo 2000). The latter require glass-ionomers, because of the need to conserve the maximum amount of tooth material, which is achieved by exploiting their inherent adhesive properties.

The biomechanical effect of tooth repair has been studied (Bharadwaj, Solomon & Parameswaran 2002), and shown to cause a significant weakening (Table 3). Repair with an adhesive system (in this case, bonded composite) mitigates this, but only to a certain extent, as shown in Table 3.

Cutting of smaller cavities and repair with naturally adhesive glass-ionomer is likely to result in teeth that retain a greater proportion of their strength than those in this study. At the very least, they are unlikely to be weaker.

Minimal intervention dentistry as an approach is likely to increase in importance in the years ahead, and thus the role of glass-ionomers in this application is expected to grow.

4 CONCLUSIONS

Glass-ionomers are bioactive cements of continuing importance in contemporary dentistry. They can be used as direct repair materials, and this use is associated with the current clinical developments of ART and MI dentistry. Modern glass-ionomer materials have improved mechanical and biological properties. Initial engineering properties often improve significantly on maturation, though this feature is difficult to study experimentally. However, these changes, together with their bioactivity, as manifested by reversible fluoride exchange and ion-release properties, appear to be the key to the further development and exploitation of these materials.

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Computational analysis of thermal and mechanical behaviour of FGM-based (Ti/HA) endosseous dental implants in normal and overloading conditions – *A method for designing an FGM-based innovative dental implant*

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ABSTRACT: Aim of the present paper is to give an overview of Functionally Graded Materials dental implant behaviour. The device is composed of hydroxyapatite and titanium alloy in order to best bear the mechanical and thermal loading conditions due to the normal and parafunctional activities of oral cavity.

1 INTRODUCTION

A dental implant is an artificial system implanted in the mandibular bone to replace one or more damaged natural teeth. It generally consists of an endosseous component completely inserted in the mandibular bone and an abutment, which connects the endosseous part with the oral cavity.

Dental implants made of Functionally Graded Materials (FGM) have given rise to an increasing interest due to their advantage to satisfy requirements as biocompatibility, strength, corrosion resistance, that are important challenges still now to overcome. The most promising functionally graded materials to be proposed for dental implants are a mixture of Hydroxyapatite (HA) and Titanium (Ti) with a continuous gradation of material composition along the vertical direction, from HA rich in the lower part that is inserted in the jaw bone, to Ti rich in the upper part where occlusive force is directly applied. To define the design model the normal occlusive loading as well as the overloading conditions are considered, that are the clenching and the grinding of the dental arches, due to the Bruxism.

A numerical approach is carried out to analyse the whole system, that is a 3D model of mandible portion interested by the dental implant, with the geometrical and loading boundary conditions that are able to simulate the natural conditions.

The implementation of the micro-structured material model allows to perform a parametric study of the FGM composition governed by an exponential law of the mixture distribution.

According to the clinical and biomechanical needs, the final goal of this work is to define

the best FGM composition, in terms of design parameters, to bear the above mechanical and thermal conditions.

So this work can be regarded as an example of straight integration between clinics, surgical practice and material innovation.

2 FUNCTIONALLY GRADED MATERIAL (FGM) – BASED DENTAL DEVICES

2.1 FGM for dental devices

Different types of materials have been used to produce biomedical implants. Metals in pure forms, such as titanium and a wide range of alloys whose main advantage is their high strength and their good resistance to corrosion. Non-metallic materials, such as carbon, hydroxyapatite and aluminium oxide crystal also exhibit good osseointegration properties, however their low strength and great brittleness limit their applicability.

The current dental implants composed of a single material, sometimes with a coating layer, are, however, essentially uniform in composition and structure in the longitudinal direction. Ideal biomaterials for use in dental implants need to simultaneously satisfy many requirements such as biocompatibility, strength, fatigue durability, non toxicity, corrosion resistance, and sometimes aesthetics. In addition, the required function of a dental implant varies from part to part.

A conventional dental implant with a single composition and uniform structure can not meet all of these requirements. Hydroxyapatite (HA), the principal component of bone and teeth,

exhibits excellent compatibility with bone, but its mechanical strength and reliability are too low for an artificial implant to be made of pure HA. To improve the implant success rate, metallic implants with a HA coating layer have been introduced and fabricated to make use of the merits of both metal and ceramic materials. Unfortunately, HA-coated implants are still uniform structures in the longitudinal direction, and the HA coating layer is often mechanically broken down or absorbed in the bone tissue [1].

The concept of FGM may be suitable to apply for implant. In Figure 1 are shown the expected properties of functionally graded dental implant. The most commonly used functionally graded bio-material (FGBM) dental implants are a mixture of HA and biocompatible metal titanium (Ti), denoted as Ti/HA, with a continuous gradient in the material composition.

Normally, its material profile changes along the longitudinal direction from Ti rich in the one end (left in the figure) where occlusive force is directly applied to HA rich in the other (right), that is implanted inside the jaw bone to achieve the optimal mechanical properties and biocompatibility.

Hence, it is important that a three dimensional model, incorporating the interaction between the implant and the surrounding bone, be adopted to provide a more accurate interpretation of the stress and displacement distributions in the

implant and the periimplant bone tissues, taking into account the very severe loading conditions due to bruxism and the thermal action due to the hosting environment.

The Ti/HA based FGM can be considered a composite material and for its components it is deemed to be a bioceramic material that is the most promising class in the available ones used for human body-implants [2]. Few of the bioceramics have a similar structure as the mineral part of our bone.

2.2 Mechanical properties of FGM materials: Mixture rule of FGM

For FG dental implant studied in the present work, Figure 2, the exponential law applied along the length of implant is chosen assuming that the value of the volume fraction of the HA component, in the dental implant, ranges from one at the apex to zero at the connection with the abutment without physical discontinuity. In this way, there will a desirable highest biocompatibility at the apex and a good strength around the abutment. The volume fraction for the HA, V_h , is expressed as follows

$$V_h = \left(\frac{H - z}{H} \right)^m \quad (1)$$

where z is the coordinate that represents the distance from the apex and H is the length of the dental implant. The exponential index m is the design variable whose best value depends on the parametric study that consists of numerically analysing the mechanical effects of different material mixture distributions corresponding to m varying from 0.1 to 10. Such mechanical effects in terms of von Mises and min/MAX stresses are interpreted in relation to mechanical resistance to external actions and to biocompatibility of surrounding bony tissues [3].

The volumetric fraction for the Ti is then written as:

$$V_t = 1 - V_h \quad (2)$$

The volumetric fraction variation along the length of the dental implant with regard to different values of m is graphically depicted in Figure 3.

The mechanical properties for the FGM implant, namely the Young's modulus E and the Poisson's ratio ν , and the thermal coefficients expansion CTE , are derived from properties of the respective volumetric fractions of the component HA and Ti as follows [3].

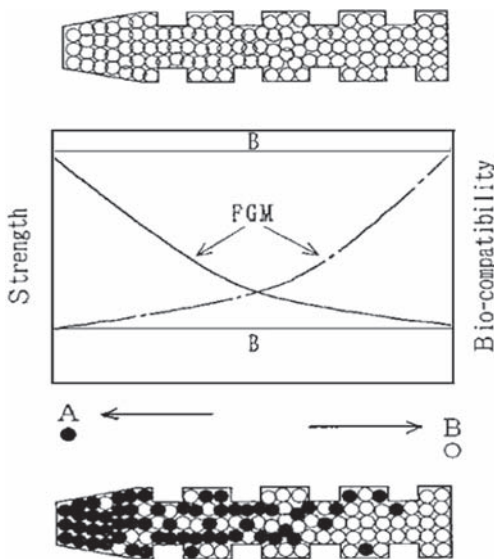


Figure 1. Expected properties of functionally graded dental implant [2].

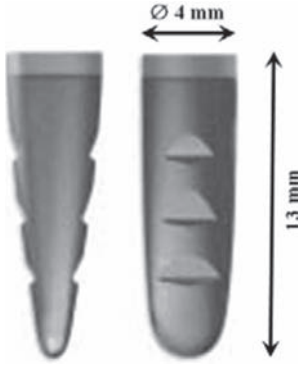


Figure 2. Frontal and lateral view of dental implant model geometry.

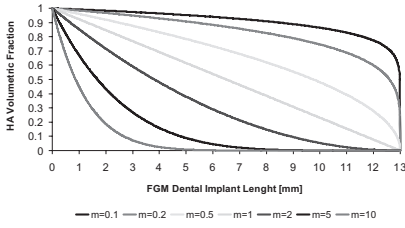


Figure 3. HA volumetric fraction along the FGM dental implant length.

$$E = \frac{E_h \{E_h + (E_t - E_h)V_t^{2/3}\}}{E_h + (E_t - E_h)(V_t^{2/3} - V_t)} \quad (4)$$

$$v = v_t V_t + v_h V_h \quad (5)$$

$$C = \frac{C_t K_t V_t + C_h K_h V_h}{K_t V_t + K_h V_h} \quad (6)$$

$$K_t = \frac{E_t}{2(1-v_t)}, K_h = \frac{E_h}{2(1-v_h)} \quad (7)$$

where subscript indexes t and h stand for Ti and HA, respectively. The calculated variable Young's modulus and the Coefficient of Thermal Expansion (CTE) with regard to different exponential index m are plotted in Figures 4 and 5.

3 EXPERIMENTAL AND ANALYTICAL ASSESSED DATA

3.1 Dental biomechanics: Normal and overloading conditions

The stresses developed in the biological structure under loading have an important role in the

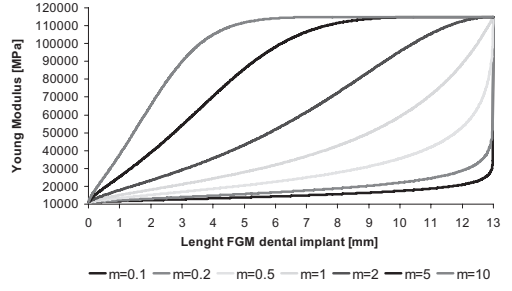


Figure 4. Young's modulus along the FGM dental implant length.

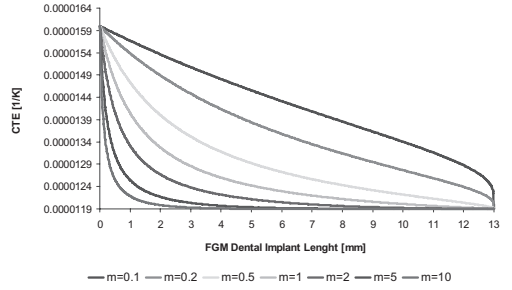


Figure 5. Coefficient of Thermal Expansion (CTE) along the FGM dental implant length.

longevity of the teeth. Stresses depend on the applied loads and the resisting contact area.

As vector quantities, their direction is an important aspect to be taken into consideration. The applied forces mainly are in axial direction, but also the horizontal components can grow being responsible for trauma leading to the loss of teeth. Abnormal or parafunctional forces also give rise to bending effects, which are proportional to crown height.

Occlusion loads are the forces involved in the process of breaking the food into smaller portions, which will be later triturated during the mastication.

According to Anusavice [4], the average maximum occlusion force is around 756 N.

Mastication loads are the forces developed between antagonistic teeth during the process of food mastication. Such forces are mainly directed perpendicularly to the occlusion plane in the posterior region of the mouth.

In order to measure the mastication forces, Anderson [5] used a strain gauge on a lower molar tooth.

Four pieces of biscuits were chewed bilaterally at eleven seconds intervals. The maximum forces measured varied from 39 to 59 N.

The definition of an ideal occlusion in clinical dentistry generally specifies bilateral and simultaneous contacts between teeth in the intercuspal position, resulting in a balanced distribution of forces.

When a contact occurs outside this ideal situation, load transmission points are changed thus modifying the stress and strain distribution on the teeth.

Bruxism is one of the possible causes for this kind of malocclusion. Defined as a frequent teeth clenching and grinding at times and for purposes other than mastication, bruxism can occur when the person is either awake or asleep, although it is more common during sleep with a frequency and intensity that depend on several neurological and physiological factors, dreaming activity included. The bite force was measured bilaterally between the first molars. According to Van der Bilt *et al.* [6] was observed a significantly lower muscle activity for unilateral clenching as compared with bilateral clenching.

3.2 Numerical modelling

The starting point was to develop an accurate model of an edentulous mandible, which is essential for obtaining more precise results.

Initially, computerized tomography (CT) of an actual human mandible and a real existing implant were obtained according to the description of Inou and co-workers [7]. With the help of a scanner, the images obtained were converted into digital data and transferred to a CAD program (AutoCAD; Autodesk, San Rafael, CA), where the coordinates of the contouring points were extracted from these plots and joined to form partial volumes that together defined the final geometry, as depicted in Figure 6.

The kinematic conditions that ensure any whole motion restriction for the entire system, are the fully constraints on the two end surfaces. Therefore any growing constraint reaction force affecting the more interesting zone in the presence of the dental implant is avoided.

The dental implant, in turn is supposed to be integrated by means of the osseous tissue surrounding the device. The biomechanical properties of the interfacial structure (or compact bone) are not well known yet, and would require suitable tests as before mentioned.

As previously mentioned, the implant is considered completely osseointegrated, and in biological terms it implies the perfect implant anchorage to mandible by means of the surrounding compact bone forming.

Extracting the trabecular part from the model of bone with the implant that is supposed to be divided into two parts, it is possible to visualize the

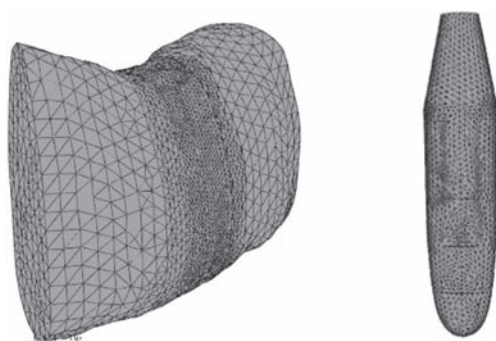


Figure 6. Mesh of the mandible portion and of the implant.



Figure 7. Model of half bone with the implant, supposed to be divided in two parts without the inner part of the trabecular bone.

cortical bone and the compact part, 1 mm thick, surrounding the medical device, Figure 7.

Besides the normal loading also the aspect of the parafunction loads due to the bruxism pathology is taken into account. For the normal loading, the value of 100 N directly applied on the top of the abutment is considered; the parafunction load suggests higher values. In fact 790 N is the vertical loading, Figure 8, on the abutment due to the clenching and the transversal force of 50 N is the grinding during the bruxism events in the rest of night hours.

Therefore these biomechanical results are extracted from the peri-implant values of the Maximum and the Minimum of the Principal Stresses (S1 and S3) as well as the von Mises stresses for the implant material strength.

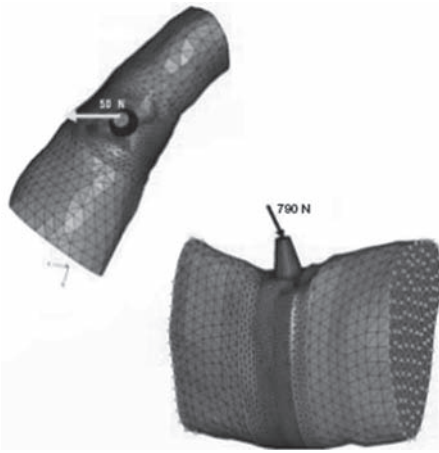


Figure 8. Parafuction Loading: grinding and clenching forces in case of Bruxism event.

4 RESULTS AND DISCUSSION

On the basis of the results that have been assessed by means the displacement-based commercial code ABAQUS (Abaqus Inc., 3DS Simulia, USA), in the present section are shown the mechanical and thermal stresses of the entire model of the dental implant with the surrounding mandible bone portion.

In particular are taken into account the thermal loading due to changing temperature of oral cavity and the mechanical actions in normal and overloading conditions due to the bruxism pathology. With more detail are plotted the von Mises, the Maximum and Minimal Principal Stresses (S1 and S3) surveyed on the interface between the hosting bone and the medical device in order to assess the reliability of the FG material.

4.1 Thermal and mechanical results

Allowing for the oral temperature variations, the thermal stresses are also calculated through the temperature load, and a variation of temperature $\Delta T = -20^\circ\text{C}$ and $+20^\circ\text{C}$ respect the ordinary oral thermal condition of 37°C for the stomatognathic system is assigned as well as the case of $\Delta T = 0^\circ\text{C}$ is considered.

In bruxism event during the nocturnal rest hours, the oral temperature is supposed to be constant and stable to the value of 37°C , so in this case it is only applied the mechanical load, Figure 9.

4.1.1 Mechanical loading

The different FG implants give very similar variation patterns of peri-implant stresses. In view of

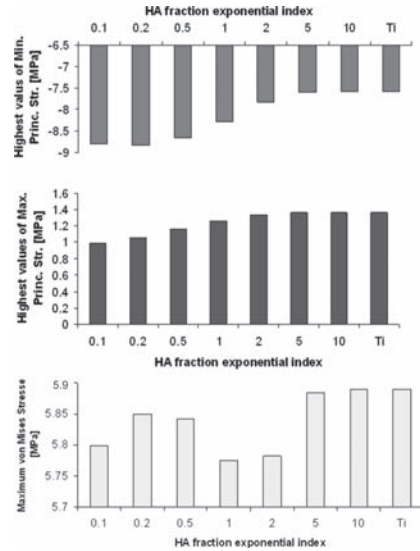


Figure 9. Histogram of maximum peri-implant stresses in buccal-lingual and mesial-distal section under occlusal force only.

the general stress level, the FG implant with the smallest HA volumetric fraction exponential index ($m=0.1$) defines the lower bound, whereas the titanium implant registers the upper bound. Is evident that the ascending order follows the exponential index increasing order, and it means that the higher the exponential index, the higher the overall Young's modulus.

Thus it can be stated that under occlusal force only, the mechanical performance of all the FG implants are almost equally good, and the titanium device sustains the much higher maximum von Mises stresses.

4.1.2 Thermal and mechanical loading

Aim of this section is to show the effect of temperature change due to daily oral cavity behaviour by means of simulations where $\Delta T = -20^\circ\text{C}$ and $+20^\circ\text{C}$ (respect the ordinary mouth temperature of 37°C) is considered, in addition to the occlusal force already shown.

In summary, the decrease of temperature increases the stresses, especially the tensile stress remarkably.

Relative mechanical performances of the FGM and titanium implants under both the temperature load and occlusal force are very different from those under occlusal force only, Figure 10.

4.1.3 Bruxism loading

In the extreme mechanical conditions due to bruxism event, the mechanical loading is around 800 N

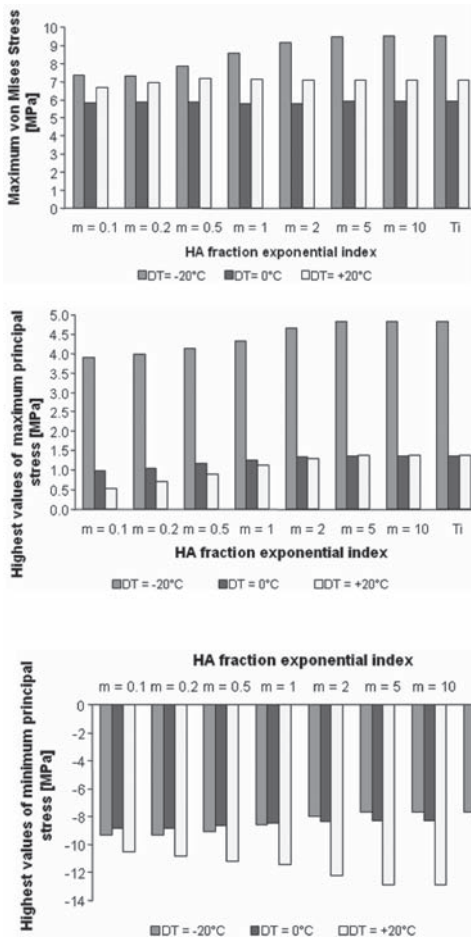


Figure 10. Histogram of von Mises, Maximum and Minimum Principal Stresses on buccal-lingual and mesial-distal section under occlusal force and oral cavity changing temperature effect ($\Delta T = +20^\circ C$, $-20^\circ C$ and $0^\circ C$).

for the vertical force and 50 N for the grinding action (transversal force). The thermal effect in this case is neglected, because the bruxism is basically an event that occurs during the nocturnal rest hours, therefore the oral cavity temperature is $37^\circ C$ with no change ($\Delta T = 0^\circ C$).

By means of the histograms in Figure 11, it is shown that all the trends for the stress level values are increased of almost one order of magnitude respect to normal loading conditions.

4.1.4 Discussion

When temperature change effect is considered, the FG implant with fraction exponential index $m = 0.2$ sustains the lowest von Mises and tensile stresses among all FG implants, whereas $m = 5$ sustains the less compressive action due to S3, thus

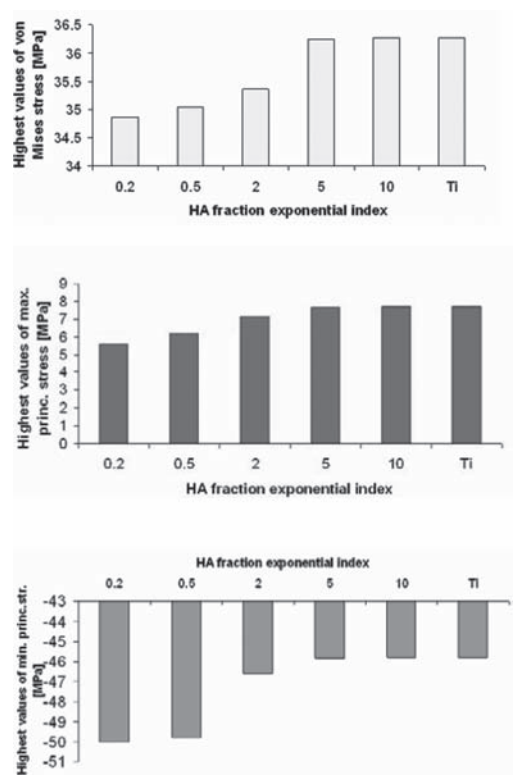


Figure 11. Histogram of von Mises, Maximum and Minimum Principal Stresses on buccal-lingual and mesial-distal section under bruxism loading. No changing temperature effect ($\Delta T = 0^\circ C$).

the thermal effect should not be ignored to evaluate the performance of this dental implant.

The normal biomechanical condition, show that exponential index $m = 5$ is the one that better sustains the compressive actions and might withstand the biological conditions.

Similar considerations hold for the bruxism loading.

5 CONCLUSIONS

For the future developments it would be desirable to focus on a general approach of FGM's in biomedical applications. The introduction of functionally graded materials in different biomedical applications can be advantageous. However, more measurements on metal-ceramic composites need to be done. These measurements include mechanical tests as well as *in vitro* and *in vivo* biological tests. With these measurements new material models specifically aiming to interpolate a variety of parameters can be developed. These material

models can then be used as input for more reliable simulation tools thus making easier the design of an FGM component.

For the specific case of titanium-hydroxyapatite in dental implants, it would be interesting to obtain long-term mechanical reliability and osseointegration in the mandible.

Allowing for the stresses in the intermediate region between the bone and implant (and therefore the suitable FGM composition) and considering the biomechanical/clinical needings, it would be desirable characterize the best available manufacturing technology that allows to produce this implant.

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Metal ceramic fixed partial denture – fracture resistance

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ABSTRACT: Metal ceramic Fixed Partial Dentures (FPD) are suitable to increase fracture resistance presenting higher clinical longevity. This type of prosthesis is mainly used when a great number of teeth replacements are needed. The FPD under analysis is defined by a metallic infrastructure (titanium) and by a ceramic coating. The advantages of hybrid FPD lie in their predictable biomechanical behaviour, versatility and cost. The main disadvantage is related to aesthetic functionality. Karlsson (1986), Lindquist & Karlsson (1998) and Palmqvist (1993) quantified the life time for hybrid FPD, referring 10 years in service to be a survival of break point. The connector design is of great importance to improve smooth stress pattern in the region between teeth. This region is also restrained by biological and aesthetic reasons. Ceramic material presents elevated failure rate in FPD due to brittleness. This work intends to predict fracture resistance to different loading conditions, using a smeared fracture approach (continuous damage mechanics). Results agree well with experimental evidence.

1 INTRODUCTION

Despite the increase of all-ceramic fixed partial dentures, metal ceramic units continue to be used due to their clinical durability and biocompatibility. Ceramic fractures represent serious and costly problems in dentistry. Moreover, they pose an aesthetic and functional dilemma both for the patient and the dentist, Özcan (2003).

Considering the existence of two or more different materials, with different biomechanical properties (thermal and mechanical) and also the adherence between them (bond strength), it is expectable to foresee problems under clinical conditions.

Failure of the restoration is dependent on different several factors. Optimum clinical design should require knowledge of failure mechanism. Besides the previous mentioned factors affecting failure, adverse environmental conditions, such as moisture and other fluids may also contribute to decrease life of FPD. The presence of microcracks at surface should be the most important reason for ceramic failure, besides the existence of pores inside ceramic material.

This paper intends to analyze the brittle behaviour of ceramic material used on fixed partial dentures, using the concept of continuous damage

mechanics. In this concept, the smear of a crack or crush is predicted by the stress level determined by tension or compression, maintaining the continuity of the displacement field where the material became ineffective.

A three unit FPD consisting of two piles and a supported tooth is analyzed, when subjected to three different loading conditions over the pontic area on the top region of crown (L1—load type one considered as two point load at the cusps zone, L2—load type two considered as ring load at the top zone and L3—load type three considered as one point load located in the fossa zone), see Figure 1.

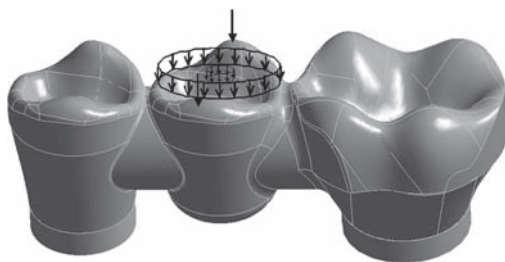


Figure 1. Three unit FPD and loading conditions, (L1, L2, L3).

All loads were applied orthogonal to the occlusal plane, using incremental procedure to predict smeared cracking and crushing.

The three unit FPD is made of a metallic infra-structure (titanium) and a ceramic coating, assuming perfect contact between them.

2 OBJECTIVES

The objective of this research is to predict damage on ceramic material, depending on load type and level. An incremental loading step was applied until the maximum load bearing was reached for each loading condition. Those different loading condition should represent a wide range of dally situations. The pattern of cracking and crushing should be determined. Cracking is the ultimate state condition under tension while crushing is represented by compressive stress state.

3 MATERIALS

Two different materials should be defined for numerical simulation of this metal-ceramic partial fixed denture. The adherence between them is not considered in this investigation, assuming perfect contact between both. The ceramic material should be considered as brittle, using adequate constitutive relations and the titanium should be considered as normal ductile metallic behaviour.

Ceramic present higher strength resistance in compression than in tension. Figure 2 represents the mechanical behaviour under uniaxial stress conditions, being the material capable of stress relieving under tension stress. This behaviour is normally used to increase numerical convergence.

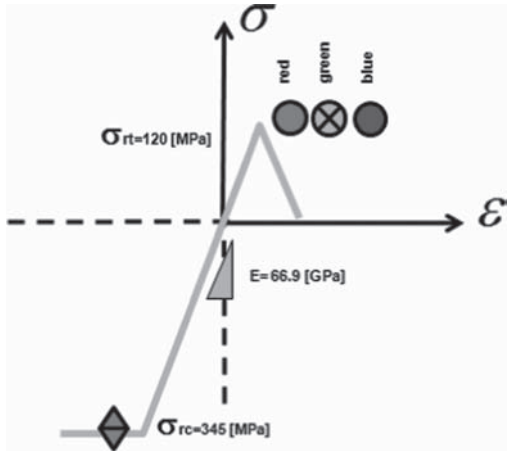


Figure 2. Stress – strain relation for ceramic material.

Material may undergo plastic behaviour under compression. Table 1 represents the material properties used together with failure mechanism, based on Willam and Warnke (1975) criteria.

Titanium alloy is considered as ductile material, which means that material presents linear elastic and may undergo plastic deformation, under tension and compression, see Figure 3. Strain values for ultimate stress may present values close to 20%.

Table 2 represents the mechanical material properties for tension and compression of titanium.

Table 1. Material model for ceramic material.

Model	Property/Function	Value
Linear (tension/compression)	Elastic modulus	66.9 [GPa]
	Poisson coefficient	0.29
Non-linear (compression)	Strain	Stress
	0	0
	0.005156	345 [MPa]
Failure model	0.010000	345 [MPa]
	Shear transfer coef. (open crack)	0.25
	Shear transfer coef. (closed crack)	0.90
	Tensile cracking stress	120 [MPa]
	Compressive crushing stress.	345 [MPa]
	Stiffness mult. for cracked tensile	1

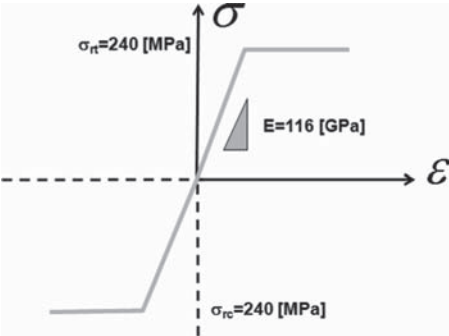


Figure 3. Stress – strain relation for titanium.

Table 2. Material model for titanium alloy material.

Model	Property/Function	Value
Linear (ten. / compr.)	Elastic modulus	116 [GPa]
	Poisson coefficient	0.34
Non-linear (tension / compression)	Strain	Stress
	0	0
	0.002068	240 [MPa]
	0.200000	240 [MPa]

4 METHODS OF ANALYSIS

The geometry of this fixed partial denture was defined as parasolid format in Solidworks CAD software and then fully transferred to the analysis ANSYS software. The geometry is mathematically modified to finite solid 65 and solid 45 elements to represent ceramic and metallic material, respectively, see Figure 4.

The metal infrastructure is a bridge in cantilever supporting condition.

Solid 65 is a three dimensional finite element with eight nodes and eight integration points, with three degrees of freedom at each node (translations in the nodal x, y, and z directions). The most important feature of this element is that it can represent both linear and non-linear behaviour of the ceramics. For the linear stage, the ceramics is assumed to be an isotropic material up to cracking. For the non-linear part, the ceramics may undergo plasticity. Cracking may take place up to three orthogonal directions at each integration point. A crack may be developed in one plane and if subsequent tangential stress to the crack face are large enough, a second (or third) crack may also be developed (red, green and blue color circle outline), see Figure 5. If the crack has opened and then closed, the circle outline will have an X through it.

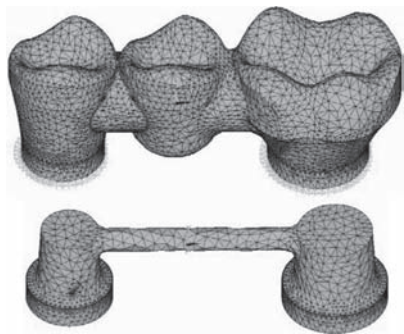


Figure 4. Unstructured finite element mesh with tetrahedrons. Complete mesh and metallic infrastructure.

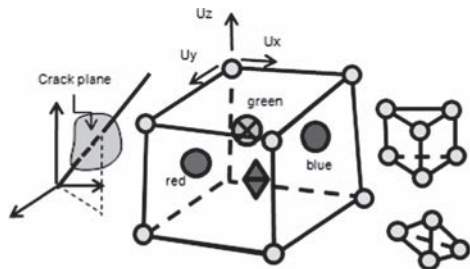


Figure 5. Finite solid 65 element.

Cracking is assumed to be spatially distributed over entire volume of element or volume attached to each integration point. The presence of a crack at an integration point is represented through modification of the stress-strain relations by introducing a plane of weakness in a direction normal to the crack face.

If the material fails at an integration point during uniaxial, biaxial, or triaxial compression, the material is assumed to crush at that point. In solid 65, crushing is defined as the complete deterioration of the structural integrity of the material and represented by an octahedron outline. Under conditions where crushing has occurred, material strength is assumed to have degraded to an extent such that the contribution to the stiffness of an element, at the integration point in question, can be ignored.

Solid 45 is a three dimensional finite element with almost the same characteristics as mentioned except for predicting cracking and crushing.

5 RESULTS

Load bearing resistance was determined for each loading condition. Table 3 resumes the ultimate load for support equilibrium. After that load level it is no longer possible to sustain equilibrium and the 3 unit FPD is considered damaged.

Progressive degradation lead to crack initiation and growth, as represented in Figures 6–8.

For load case L1, cracking is initiated next to the loading zone (cusps) and progressive damage also stars at the bottom of the abutments, in the neighbourhood to the bottom ceramic material.

The stress field is strongly dependent on fracture prediction, because material is losing resistance near cracks and crushed ceramic material.

For Load case L2, cracking is initiated next to the left abutment with progressive damage in the neighbourhood to the bottom ceramic material.

The stress field is similar to the resultant stress field for load case L1, and is also dependent on fracture progressive damage.

For Load case L3, cracking in initiated at the bottom of the abutments, with progressive collapse in the loading zone and also in the connecting bridge element, which represents traditional collapsing mode, as reported by different authors, Tsumita et al. (2007).

Table 3. Fracture resistance.

Loading	Ultimate load
L1	128 [N]
L2	201 [N]
L3	514 [N]

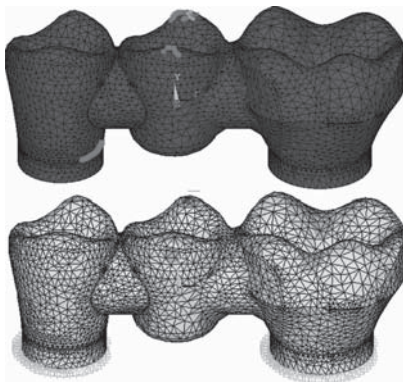


Figure 6. Ultimate limit state condition of FPD to load type 1 – LT1. Fracture prediction and longitudinal stress field.

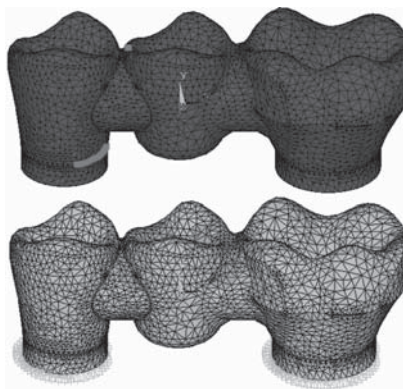


Figure 7. Ultimate limit state condition of FPD to load type 2 – LT2. Fracture prediction and longitudinal stress field.

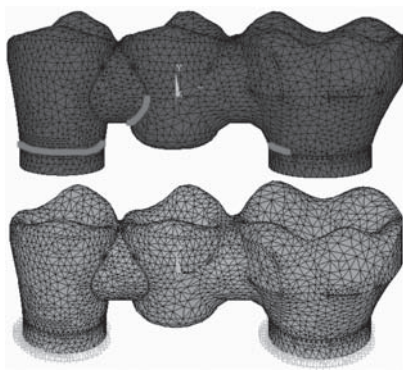


Figure 8. Ultimate limit state condition of FPD to load type 3 – LT3. Fracture prediction and longitudinal stress field.

The stress field presents two well defined compressive and tensile zones.

The connection bridge presents different design near central tooth (pontic). The rounded shape on the right contrast with the sharp geometry on the left, which is responsible for increasing the stress level and simultaneously with progressive failure.

6 CONCLUSIONS

There is consistent epidemiological evidence that mechanical failure of a dental prosthesis occurs after a certain number of service years. In case of prosthodontic restoration, ceramics cannot be added intraorally due to processing conditions. Replacement of a failed fixed partial denture is not a practical solution, reason why this type of prosthesis should be carefully design for maximum life cycle, Özcan (2003).

The most frequent reasons for ceramic failures are related to progressive cracking. Sharp shape geometry should be avoided to decrease maximum stress level.

Three different loading conditions were tested, leading to different fracture resistance. Load case L1 presented smaller fracture resistance due to localized effect of the applied force. Progressive collapsing near the abutment was revealed. Load case L2 presented higher fracture resistance, but failure occurred in the same location as load L1. This is mainly due to the similar resultant stress field. Load case L3 revealed maximum fracture resistance, with typical collapsing mode.

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Pre-school telediagnosis of dental problems: A teledentistry project

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ABSTRACT: Early childhood caries is a common disease of childhood and represents a serious problem in pediatric dentistry. In Portugal, the prevalence of caries in children with only 6 years old is around 50% and most of them only begin dental visits after this age. On seeking to reverse this trend, remote diagnosis can be a procedure to explore in order to promote the appropriate early diagnosis and treatment, so that children can maintain an adequate oral health. For the purpose, a teledentistry project was developed for remotely screening dental problems based on non-invasive photograph, using accessible and low-cost technologies. Remote diagnoses were performed by different Dentists and data were compared with in-person examination (gold standard). The results suggest that this teledentistry project could constitute a valuable approach to help in the early diagnosis of dental problems acting, so that children could be referred to dental treatment and thus can sustain a healthy oral status.

1 INTRODUCTION

Early childhood caries is a common disease of childhood and represents a serious problem in pediatric dentistry. Not only because of its rapidity but also because of age of affected children which can be by the time they reach kindergarten. Although the prevalence of dental caries has been declined in the last decade due to improvements in dental technology, use of fluoride and better knowledge about prevention of dental caries, unfortunately, many parents underestimate the importance of the first dentition. The primary teeth keep not only the space for the permanent ones but are also necessary for speaking, chewing and appearance, what is important in the self-esteem of the child (Seminario & Ivancaková, 2003). In addition, this lesion may cause general health problems, pain, difficulty in chewing and the need for premature tooth extractions which can have a profound effect on dental arch status (Capela, 1994).

Moreover, various other problems, namely malocclusion problems can also frequently occur already in children between 4 to 6 years, mainly lack of space, which can also cause negative inherent consequences for the child. Thus, early attention may also be given to malocclusion (Tschill et al. 1997). This is troublesome, as primary teeth in pre-school-age children must last several years until the permanent dentition has emerged (Kopycka-Kedzierawski et al. 2008).

In Portugal, the last National Study of Oral Diseases Prevalence shows a significantly high prevalence of caries (49%) in children with only 6 years of age and a 2.10 dmf index (average number of primary teeth decayed, missing and filled), what is worrying (Portuguese Ministry of Health, 2008). The Stomatology and Dentistry Portuguese Society associated to Colgate Portugal refer similar results (50.1%) and that only 19.4% of 6 years old children presents healthy teeth (SPEMD & Colgate, 2008). Worsening the situation, another study refers that most of the children (56%) only begin dental visits after this age (SPEMD, 2007).

According to these facts it is important to take measures that make dental problems diagnosis already in pre-school age possible in order to promote the appropriate early treatment, mainly in a preventive way and from deciduous teething phase, so that children can maintain an adequate oral health. Distant diagnosis based on images could be one of these measures to explore and may be an excellent way to inform parents about oral health of their children.

Supporting this perspective, technological developments in the photographic process have continued to change and improve the practice of dentistry. Clinicians must now integrate existing photographic principles with today's contemporary camera systems and computer software technology. This evolution to a contemporary photographic process is revolutionizing the way

clinicians diagnose, treat, and communicate with patients and colleagues (Terry et al. 2008). Images can be sent to a remote location for analysis, diagnosis and referral. There are studies which refer that digital images have great potential to accurately identify oral conditions that may be used for referral and treatment recommendations, as well as for consultation among specialists and primary oral health care providers (Kopycka-Kedzierawski et al. 2008; Kopycka-Kedzierawski & Billings, 2006).

It has been predicted that teledentistry will give an opportunity to integrate dentistry into the larger health-care delivery system and quickly will become much more than a mechanism to facilitate interactions between specialists and will be the clinical dimension of the new doctor-patient relationship. (Bauer & Brown, 2001). In fact, the developing field of teledentistry has the potential for benefiting dental care by enhancing early remote diagnosis, timely treatment of oral diseases, improved utilization of dental services and access to care, and additionally, reducing time lost from work or school. In the long term, teledentistry may also help to establish a “dental home” for participating children. Teledentistry examinations can offer an efficient initial assessment of the oral health of very young children (Kopycka-Kedzierawski et al. 2008).

Thus, teledentistry could help the process of early diagnosis of dental problems, by proving an infrastructure for remote and off-line diagnosis based on sending to Dentists oral photographs of children taken in their kindergartens, preferentially using non-invasive methods for economic and asepis reasons. This proceeding will be less stressful for small children than the conventional oral examination considering that no instruments are used, the camera remains outside the mouth and the examination is conducted in a familiar environment. (Kopycka-Kedzierawski & Billings, 2006).

With this study, promoting the development of a teledentistry pilot-project, the authors aim to evaluate the validity of dental problems remote diagnosis based on non-invasive photograph, using accessible and low-cost technologies.

2 METHODS

For the purpose, a web-based system called MedQuest was developed by the Biostatistical and Medical Informatics Department of Faculty of Medicine of Oporto University. MedQuest is implemented in Apache and PHP, and stores the data on an Oracle SGBD. MedQuest allows registered users (in this case Dentists) to have access to their personal set of cases. Users

accessed MedQuest web site from home using the Internet. After logging in, each user is asked to choose from a list of cases. These cases correspond to sixty-six selected children with pre-school age that were first in-person examined in their kindergartens by an experienced Dentist under appropriate conditions of light and using a sterilized exploration kit. Three oral/dental photos of each examined child were after taken by previously trained Childhood Educators, using a Nikon Coolpix L3 digital camera with 5.10 Megapixels resolution. After selecting the case (anonymised children identifiers), a predefined questionnaire appears for the user to insert their data. The questionnaire included three photos of the child and 44 questions about the interpretation of these photos (remote diagnosis) by the Dentist (Fig. 1). These questions included the whole deciduous teething, some definite teeth, most important orthodontic problems and other general dental problems. More specifically, the questionnaire pretends to screening initial and/or advanced caries, presence of tartar, gingivitis or dental fractures. At orthodontic level, it intends to diagnosis cases of malocclusion, namely dental crowding, open bite, overjet or overbite. As a final and conclusive question, this questionnaire asks about the need to refer children to a Dentist for dental problems treatment.

Finally, we proceed to the statistical analysis, comparing the remote diagnosis data with in-person's one. MedQuest allows exporting data, a crosstab format to facilitate statistical analysis.

This teledentistry project was previously approved by a commission of ethics and the General Direction of Innovation and Curricular

Figure 1. Screen-shot of MedQuest web-based questionnaire. On the left the three photos of a particular child and on the right the partial answers of an examiner.

Table 1. Sensitivity, specificity, confidence intervals (CI), Positive Predictive value (PPV) and Negative Predictive Value (NPV). Sixty-six children observed.

Remote examiners	Sensitivity		Specificity		PPV (%)	NPV (%)
	(%)	[CI]	(%)	[CI]		
Dentist 1	100	[87,100]	76	[61,90]	80	100
Dentist 2	97	[82,100]	52	[34,69]	67	94
Dentist 3	94	[78,99]	100	[87,100]	100	94
Dentist 4	100	[87,100]	64	[47,80]	73	100
Average	98		73		80	97

Development and all the children affected by caries were referred to free dental treatment integrated on the Promotion of Oral Health National Program (PNPSO).

3 RESULTS

The collected images presented acceptable resolution, colour, brilliance and contrast and the collaboration of the children involved in the project was considered excellent.

The gold standard (in-person examination) found a prevalence of caries near to the national studies (47%) and that 50% (n = 33) of the children needed to be referred to dental problems treatment.

Regarding the remote diagnosis and when comparing to the gold standard relatively to the final question about the need to refer children for dental treatment, the results show sensitivity between 94% and 100% (98% in average) and specificity between 52% and 100% (73% in average). The positive predictive values were between 67% and 100% (80% in average) and the negative predictive value between 94% and 100% (97% in average), (see table 1).

Sensitivity and specificity evaluation on each of dental problems considered on the questionnaire will be subsequently done intending to study other agreement rates between the dentists involved on this teledentistry project.

4 CONCLUSIONS

Within the limits of the study and based on these results, we conclude that remote diagnosis of children dental problems in our population based on non-invasive photographs constitute a valid resource when we pretend to exclude referred children to a dentist for treatment of dental problems (the average negative predictive value is 97%), but further studies should be carried aiming to increase

the validity of this proceeding to referring children for the same treatment (the average positive predictive value is 80%).

We think that some reasons of the differences found can be explained by the possible difficult on seeing posteriors areas of the mouth and the different individual interpretation of the specific classification criteria considered on this project. However, the excess or lack of zeal, intrinsic of each Dentist character, may also interfere with the answer to this final and conclusive question.

Nevertheless, this method could be used to help in the early diagnosis of dental problems acting, so that children can maintain appropriate oral health. Moreover, the quality of images also suggests that the used low-cost technology will be able to constitute a resource to explore. Although cost-effectiveness studies of this method should be preformed, we feel that having the photographs taken by school teachers and using currently available technologies (e.g. low-cost digital photographic cameras), increases the feasibility of regional or national scale children oral health campaigns.

The specificity of the test would probably be improved if the dentists had some feedback on their evaluation, as they would learn from their mistakes. An evaluation of the learning curve after having feedback should be considered as future work.

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The strain patterns of the mandible for different loadings and mouth apertures

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ABSTRACT: Numerical and experimental models of the mandible biomechanics were elaborated to characterize the human temporomandibular joint aiming the development of a condyle implant. A model of the mandible was based on a polymeric replica of a human cadaveric mandible, obtained using a 3D shape acquisition equipment. The finite element model was generated and convergence tests were realized. The boundary conditions were considered for three fixed positions, with the load applied separately on the incisive, canine or molar teeth. These boundary conditions were applied for two mouth apertures (5 and 15 mm). The two condyles could slide on the plane surface of the support. In each situation the most relevant muscles were taken into account. The most severe loading case to the condyle and mandible was found for the incisive tooth support with a minimum aperture of 5 mm.

1 INTRODUCTION

There are several diseases that can affect the human mandible behaviour, among which we highlight the cancer, trauma or fracture, a congenital malformation, osteochondritis (Dingwerth et al. 2003). In the United States TMJ diseases may affect 30 million people. While a vast majority of these patients can be treated without surgery, a small group requires surgery and different implant solutions (Quinn 2003).

The pain relief and functional recovery of the joint are the most frequent causes for the achievement of the TMJ arthroplasty and other applications (Wolford et al. 2003).

The development of a new component to restore the mandible functionalities needs to reflect the loading conditions and to respond to the most severe loading situations. The functional loading conditions will cause changes to the microstructure of the bone and may possibly change the bone response (Iwasaki et al. 2003).

The mandible bone is of complex geometry and boundary conditions need to be correctly specified, otherwise these can undermine the reality of results.

Because of obvious practical reasons, a direct behavior measurement in the functioning human mandible cannot be achieved. It was suggested that finite element model may be used instead to predict real biomechanical responses using mandible models (Daegling et al. 1998).

In this sense, it is demanding that numerical models are tested and validated experimentally (DeVocht et al. 2001, Koriath et al. 1992).

Finite element and experimental models have been used to determine stresses and strains on the surface of the bone structure (Al-Sukhun 2007).

It is important to consider the muscle loads for the simulation of the mandible behavior. The directions and the magnitude of these loads have been reported using different techniques, like MRI and EMG (Mesnard 2005) or using finite element models (Tanne et al. 1993).

The objectives of the present study are to investigate and to analyze the strain pattern along the external surface of the mandible during mastication activities. A 3D numerical model of a human mandible was built and strains evaluated from generated static mastication forces on the incisive, canine and molar teeth and apertures of 5 and 15 mm.

2 MATERIALS AND METHODS

2.1 A CAD Model

The mandible model was based on a polymeric replica of a human mandible from Sawbones® manufacturer (model 1337 was selected). This model is a geometric model with necessary accuracy to be used in experimental procedures, similar to the one presented by Santis et al. 2005.

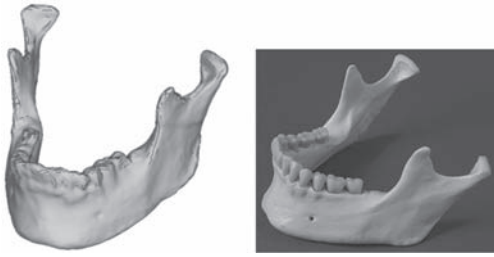


Figure 1. CAD model and a replica of a mandible.

The model geometry was obtained using a 3D laser scanning (Roland LPX 250 machine) device.

The complexity of the geometry involved the completion of ten scans with different orientations. The resolution of scan was 0.2×0.2 mm, and the final geometry is represented in Figure 1.

The final construction of the mandible was made with CAD software (CATIA, Dessault Systems). A solid homogeneous polymeric model was considered. Ichim et al. (2007) concluded that the thickness of the cortical bone does not have significant influence on strain distribution on mandible.

2.2 Finite element model

Finite element models (FEM) are important tools to determine the behavior of complex structures. The FEM used in the present study has been previously validated with an experimental procedure using strain gauges (Mesnard et al. 2006). The FEM was built with Hyperworks® 8RS1 and runs were performed with MSc MAR™ solver.

The FEM was composed of 71280 tetraedric linear elements with 4 nodes and 51245 degrees of freedom (Figure 2).

We considered that the tooth had little influence on the mandible and, particularly, on the condyle behavior. This hypothesis was validated with an experimental study (Mesnard et al. 2006) that also validated the boundary conditions applied on the FEM.

A mesh convergence test was carried out with different mesh sizes. The maximum displacements were analyzed and convergence was reached for 50000 DOF (Relvas et al. 2009). Other authors refer that convergence rate was reached for a mesh of more or less 25000 DOF for a human mandible (Hart et al. 1992).

A fixed contact point was considered for the right condyle and, the three rotations of the bone remained possible. The right condyle presented one contact point on the reference frame and two translations. The three rotations of the bone remained possible. The simulation took into account the

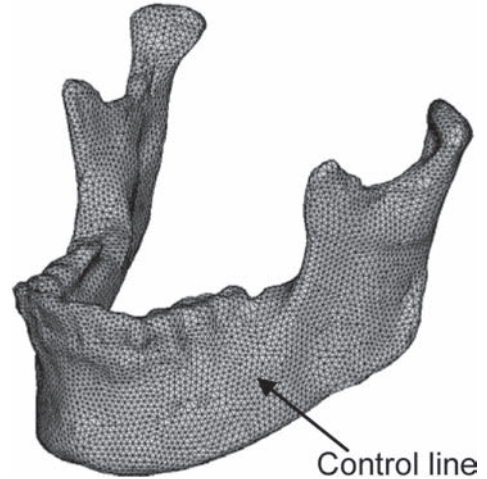


Figure 2. Mandible finite element model.

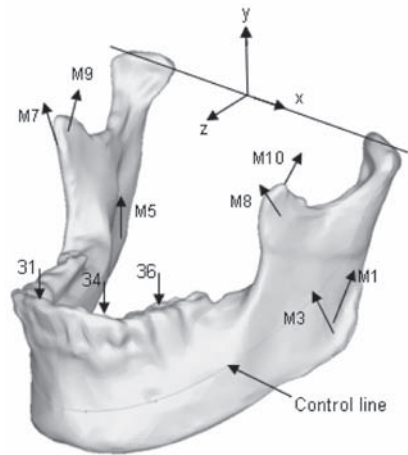


Figure 3. Position and muscular actions, M1, 2 – Deep masseter, M3, 4 – superficial masseter, M5, 6 – Medial pterygoid, M7, 8 – Temporal, M9, 10 – Medial temporal.

mandible mechanical properties supplied by Sawbones, namely a Young modulus of 260 MPa and a Poisson coefficient of 0.3. The muscles forces were applied in the same position for all models. The position was determined through MRI images (Mesnard 2006).

2.3 Load cases

To simulate the bite force, the load was applied successively on three different teeth, 31 (incisive), 34 (canine) and 36 (molar), represented in Figure 3.

Table 1. Loading configuration.

		Load(N)				
Muscle sensor		Deep	Sup.	Pterygoid	Temporal ant.	Medial temporal
		masseter max	masseter max			
31	5	129.6	303.3	286.5	15.7	9.7
34	5	92.3	215.9	272.3	115.2	53.7
36	5	81.4	208.5	382.5	133.7	64.8
31	15	129.4	228.5	363.1	37.2	24.7
34	15	129.6	228.2	364.5	63.7	38.4
36	15	216.5	259.9	146.4	146.0	91.1

Localization of the sensor position was represented and muscular forces. For each of these three positions, the two different inter incisor apertures 5 and 15 mm were considered. A sliding plane was considered on the condyle.

The loading conditions were defined based on the work of Mesnard 2005, that evaluated the muscle forces using in vivo MRI and EMG. Table 1 presents the load intensity for each muscle. The vector depends of the loaded tooth and of the aperture.

The protocol was approved by the two laboratories (Mesnard 2005) and freely accepted by the six volunteers (medical students). The study aimed to characterize the muscular actions, i.e. the forces exerted by the five elevator muscles which were considered: deep and superficial masseters, pterygoid, anterior and medial temporal. In the position 36 (Fig.3), the simulated load was similar to the load used by Ichim et al. (2007).

3 RESULTS

The results were obtained with FEA. The intact mandible was analyzed considering the influence of the sensor in 3 positions and for 2 mouth apertures.

3.1 Sensor position

Figure 4 shows the mandible displacements in the x- direction for each of the three tooth supports (sensor in position 31, 34 or 36).

On the x-axis the position was defined from the right condyle to the left and the origin was considered on the right condyle. The boundary conditions were maintained for all the simulations relatively to the muscle positions.

The positions of the condyles were maintained constant and the mandible was allowed to rotate according to the two apertures, in the three different positions. One could observe that the worst

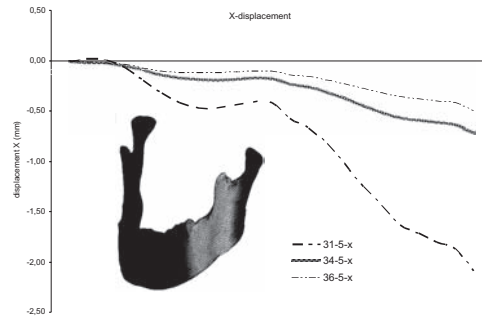


Figure 4. x-displacement for the 3 different positions.

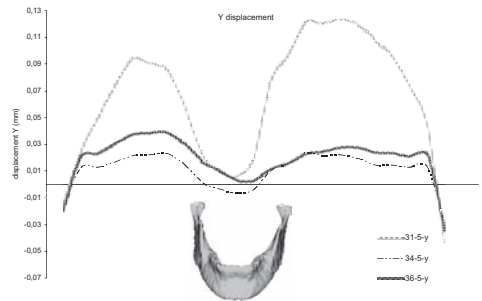


Figure 5. y-displacement for the 3 different positions.

situation corresponded to position 31 and verify that the largest x-displacements of the contact point corresponded to the position 31.

For the other two other positions, 34 and 36, the results were very similar but position 36 developed a less condyle displacement.

Relatively to the y-direction the behavior was similar on positions 34 and 36 (Figure 5). The worst mandible loading simulation was obtained at position 31, at the middle of the mandible. The displacement for this direction was 10 times less than the one in the x-direction near the condyle.

For each of the three simulations, in the z-direction, the displacements remain under 0.6 mm. The three sensor positions had no relevant influence on the results, but at position 31 the displacement was 10% higher than the one at position 36. The lowest displacement values was obtained at position 34.

3.2 Mouth aperture

For both apertures, 5 and 15 mm, the x-displacement for the three positions was analyzed. Figure 6 presents the x displacement for the same controlled points.

The x-displacement results confirm that position 31 is one of the most severe loading for both,

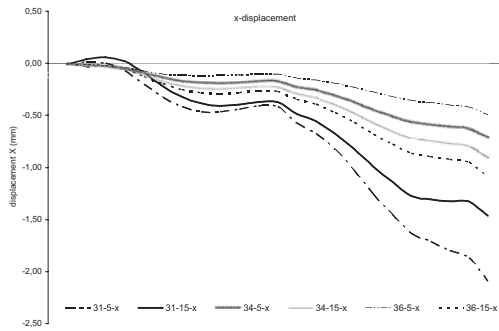


Figure 6. x-displacement for the 3 different positions and the 2 different mouth apertures.

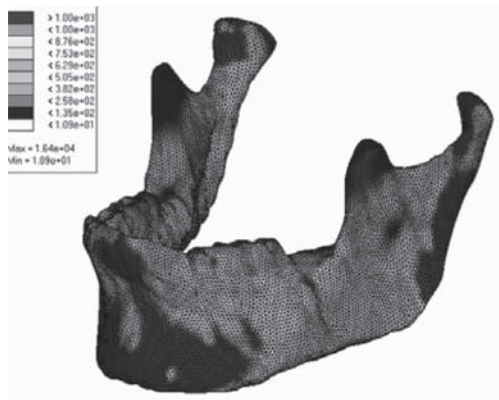


Figure 7. Equivalent Von Mises strain on position 31.

condyle and mandible. Changes occurred at positions 34 and 36, where the mouth aperture changes the behavior. For these two positions and for the larger mouth aperture the loading increases the condyle displacement.

In the y and z-directions the mouth aperture has the same influence on the displacements. The most severe loading condition was obtained for position 31 for the minimum aperture (5 mm). The influence of the loading on the condyle was only critical for the x-displacement due to the imposed boundary conditions on the condyle (sliding plane).

4 DISCUSSION

One can observe that the more severe loading situation relatively to the displacement was for position 31 for a mouth aperture of 5 mm. Therefore we study other apertures values for position 31

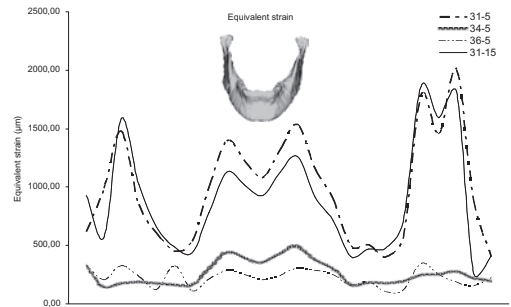


Figure 8. Equivalent von Mises strains for the 3 different positions and 5 mm mouth aperture.

and verified that the result for 5 mm aperture was more severe than others (15, 20 and 30 mm).

To confirm this fact we analyzed the deformation patterns on the mandible. Figure 7 presents the results on the mandible for position 31 and 5 mm mouth aperture. The results demonstrate that the critical zones are situated near the condyle and on the front of the mandible. These regions are more cortical because of the bone thickness of the natural mandible, and that confirms the accuracy of the FEM.

To verify the mandible deformation patterns we analyzed on the same control points the equivalent deformations. Figure 8 represents the deformation of mandible for three positions (31, 34, and 36) and for 5 mm mouth aperture.

The results show that position 31 was the most critical for the deformation near the condyles and on the front of the mandible. Perhaps yet almost said before Figure 7. The numerical results were almost symmetric. The results for position 31 and for 5 and 15 mm aperture verify the influence of the mouth aperture. Similar displacements were obtained in position 31.

5 CONCLUSIONS

The study here presented demonstrates the influence of the boundary conditions over the mandible behavior.

The results at position 31 (support on the incisive tooth) were the most critical relatively to the three simulations (load applied on the canine and molar teeth).

The mouth aperture was analyzed and the minimum aperture (5 mm) created the most severe loading at position 31. However, position 34 and 36 were more critical for the 15 mm aperture. This study can be useful for the design of biomechanical systems and for the analyses of implant joints.

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Maxilla bone pre-surgical evaluation aided by 3D models obtained by Rapid Prototyping

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ABSTRACT: For implant planning and placement, the association of CAD and CAM techniques furnishes some advantages, regarding 3D determination of the patient's jaw anatomy and fabrication of both anatomical models and surgical guides (Verstreken et al. 1996). In this paper we will present Rapid Prototype (RP) use as a tool, able to produce solid models of a maxilla in order to allow pre-surgical conditions evaluation in a patient who has lost bone tissue and needs dental implants. In this process, three-dimensional reconstruction has been made from a bi-dimensional image file, obtained by Computerized Tomography (CT) and a set of partial and total biomedical models have been manufactured to allow maxilla analysis. Rapid Prototyping technique used has been three-dimensional printing (TDP or 3DP) which allows a good reality simulation.

1 INTRODUCTION

In this study we will present Rapid Prototyping (RP) use as a tool to manufacture a biomedical 3D model from a human maxilla for pre-surgical study procedures.

The patient has suffered bone tissue loss and is intended to evaluate if it is possible and which are the best maxilla regions to promote bone tissue regeneration in order to obtain the needed thickness to apply the dental implants.

It is meant to do 3D reconstruction of a biomedical model from a 2D image file obtained from Computerized Tomography (CT) scan. After this reconstruction it is used Rapid Prototyping technology—Three Dimensional Printing (3DP or TDP) to produce the solid model.

By evaluating the conditions presented from the 3D model a previous surgery plan will be performed and presented to the patient to ensure that he is aware of his condition and the needed procedures that will lead to his treatment.

1.1 Maxilla bone tissue loss

With teeth loss the stimulation that allows alveolar bone to keep growing and maintaining, disappear,

leaving place to a degenerative process that makes, in the beginning, bone edge stretching and trabeculae decrease and for last, global bone height shortening (Cardoso et al. 2002).

Posterior maxilla has a poor bone quantity and volume, reduced by pneumatization of maxillary sinus (McCarthy et al. 2004). To oppose that anatomical limitation, maxillary sinus elevation has been one of the common surgical procedures in dental implants treatment (Mazor et al. 1999).

Requisites for the implant osseointegration success are the reconstruction material, implant design and surface finishing, surgical technique, reception place and charge conditions.

As example of used reconstruction material we can refer the allogene bone (like bovine bone), alloplastic materials (usually hydroxypatite) and autogene bone that can be obtained from extrabuccal areas like iliac bone or cranial calota or from intrabuccal areas like the tuberosity and the mentonian and back molar area. However, autogene bone is the one that presents the nearest pattern from the ideal to buccal bone reconstruction (Graziani et al. 2004).

Maxillary sinus elevation procedure can be made in two steps with a cicatrization phase from about 4 to 6 months in order to allow biological

integration of the added material (Schlegel et al. 2003). After this, implants are placed in position.

Depending of the reminiscent alveolar bone quantity, implants collocation can be made, simultaneously, at same time than bone addition for maxillary sinus elevation (Sendyk & Sendyk 2002) (Wragg et al. 2004), having as advantage in surgical phase the cicatrization period shortening, less surgeries and a lower risk of added bone re-absorption (Chiapasco & Ronchi 1994) (Misch 2000) once the implant, it self, acts like a support. However, this technique is only possible in cases where can be obtained a good primary stability, usually in situations where there is, at least, 5 mm of bone.

Used techniques for maxillary sinus elevation, usually, include a reception place enlargement using a wide variety of materials. The most used technique is the one that allows lateral access (windows), in which implants can be placed in areas considered inadequate by the insufficient bone anatomy (Avera et al. 1997). In cases where only some bone atrophy can be registered there are used osteotomes through an access in definitive place implants position, gradually expanding the bone and apically displacing it in the maxillary sinus to obtain a localized elevation between 2 and 7 mm (Toffler, 2004), being this one the less invasive technique.

1.2 *CT images conversion to 3D models*

In the conversion process of a computerized tomography in to a 3D model, it is needed a sequence of cross sections from the study object. Using 3D reconstruction software it is possible to transform these bi-dimensional images in a three-dimensional model that can be used to produce a solid model in rapid prototyping equipment (Foggiatto 2006).

Images obtained from computerized tomography obey to the international standards from DICOM (Digital Imaging and Communications in Medicine) pattern. Those are obtained from axial cuts of the study area and the equipment should be settled to the less possible thickness, once the lower this value is, better will be the model quality (Foggiatto 2006).

1.3 *Rapid prototyping and some medical applications*

Rapid Prototyping is the automated manufacture of physical objects. It is an additive-constructive process, layer by layer that allows complex form objects direct production from three-dimensional data and that is used to manufacture solid prototypes (Rocha & Alves 2000). The geometries needed can be obtained using some CAD software or through the conversion of data proceeding

from 3D Scanners, Computerized Tomography or Magnetic Resonance devices. The first techniques to Rapid prototyping become available in the eighties and were used to produce models and prototypes parts (Alves & Braga 2001).

One of the main applications of Rapid Prototyping is the fast way that is allowed in verifying new concept projects, when those are in the earlier stages or even in advanced phases of conception. In all Rapid Prototyping processes is used a 3D CAD model that is translated into an STL (Stereolithography) format file, (Souza et al. 2003) where all the model surfaces are converted in a triangle mesh.

In Biomedical Engineering field, using Rapid Prototyping techniques it is possible to produce several types of anatomical models and implant replica with educational purposes or to better understand a specific patient pathology. The models, depending of available techniques, can be made of paper, wax, ceramic, plastic or metal (Antas & Lino 2008). These models can be produced without finishing or color or have these finish operations done later to improve visualization. With educational purpose it is possible to manufacture implant replica with much lower cost than the implant value.

A great interest can be found in anatomical models manufacture from patient tomographic images. These models allow students from biomedical field to have an easier view of a specific pathology and compare it with normal anatomical models. To better understand image techniques and anatomy, it is also possible to compare, simultaneously, the original image (TC or MRI) and 3D solid model.

Medical professionals have cooperated with other field professionals in a way to optimize pre-surgical pathology analysis, shorten surgical times, create personalized tools, turn easier the communication with patients and, simultaneously, to explore the capabilities that this technology offers in personalized prosthesis design (Antas & Lino 2008).

Several manufacturing processes are available today, as Fused Deposition Modeling (FDM), Stereolithography (SLA), Selective Laser Sintering (SLS), Tridimensional Printing (TDP or 3DP) and Laminated Object Manufacturing (LOM) among other specific processes.

2 METHODOLOGY

After patient's authorization for TC images be used, these were transferred to the computer where would be done image processing and removed all personal information data.

The process to obtain the anatomic model is composed by the following steps:

- Pre-processing from bi-dimensional images and reconstruction from the surface between the contours is done in image processing software ScanIP®. This step is done by using image processing operation such as threshold, *floodfill* and *paint*, which allow the creation and distinction of the masks, based in image grey levels. These masks can be defined through the color choice done by user allowing giving the desired contrast degree to the model for an easier visualization as well to enlighten the desired elements.

First step of conversion consisted in 3D representation through the image processing application that allow closed volume visualization, after a segmentation operation based in the signal intensity—*thresholding*. This interactive application allow the user to detect and select contours in the maxilla area by doing a redefinition of grey levels that led to a separation of the bone from soft tissues. This operation applies gray levels recognition algorithms allowing, this way, a bigger grade of automatization.

After obtaining the contours with the desired quality, those are enhanced in a manual way using *paint* and *floodfill* operations. These operations consist in adjusting the obtained contours to the shape of the elements intended to represent and model. This step reveals it self as the most time consuming once the contours should be adjusted in more than one orientation (with axis changes) and in a manual way in each image to be processed.

- Rendering and 3D visualization that allow following work development during the previous step, to detect and correct possible imperfections. 3D rendering is done by the application of a consecutive planar triangle mesh from the masks defined in previous steps. Combining these two last phases it is possible to do an iterative process with the objective of present the model as close to the reality as possible.

In figure 1 it is possible to see model imperfection in a phase previous to manual masks adjustment.

In this case, not only a complete maxilla model have been produced, but also a global model divided in five transversal sections with the aim to allow a better view of the width of cortical bone along the maxilla.

- STL (Stereolitography) data generation allows combining all the active masks in a single file or the creation of several files with distinct masks.

Contained data in this kind of files consist in the conversion and translation from the 3D model mesh outputted from image processing software in

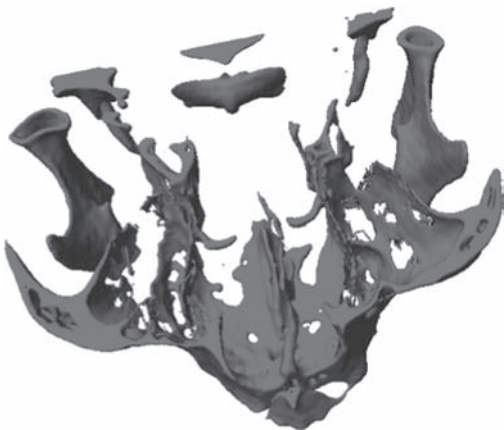


Figure 1. Image pre-processing done in ScanIP®.

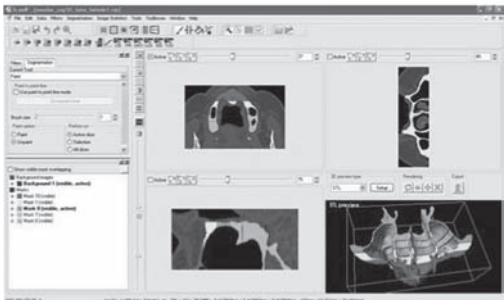
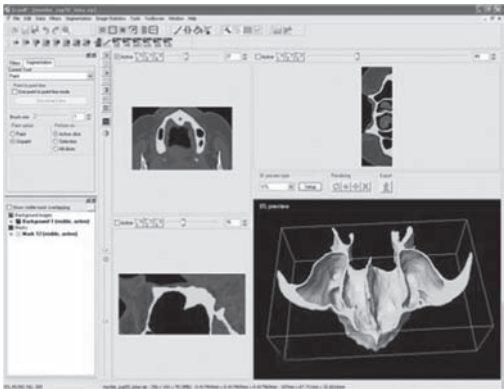


Figure 2. Images pre-processing done in ScanIP® software.

to a printing format recognized by the rapid prototyping device. This format contains the model layer division in a way to allow the layer by layer printing.

- Model manufacturing in Rapid Prototyping device Zprinter 310 from ZCorp.

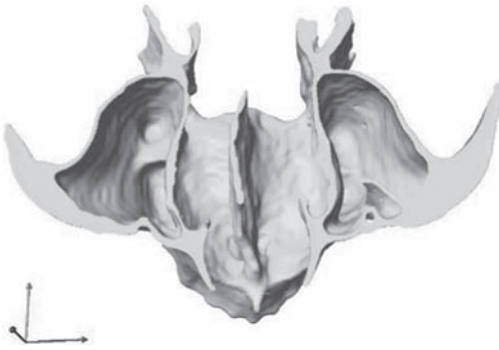


Figure 3. Rendering and 3D previewing of maxilla total model in ScanIP® software.



Figure 4. Rendering and 3D previewing of maxilla transversal sections in ScanIP® software.

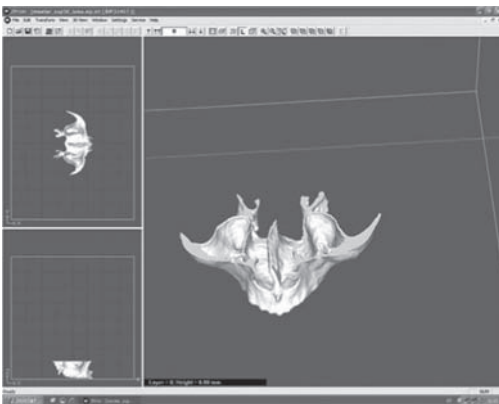


Figure 5. STL file visualization in printing software ZPrint®.

In the images from figure 6 it possible to visualize several phases from model manufacturing, going from the layer impression until the cleaning of the residual dust.

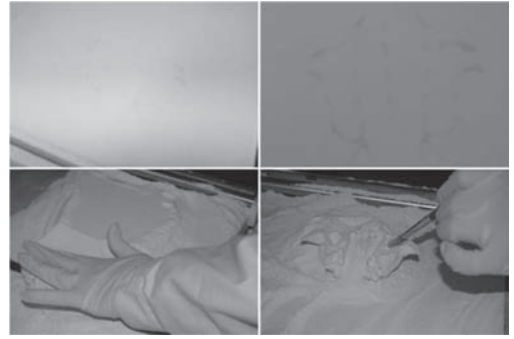


Figure 6. Complete model manufacturing.

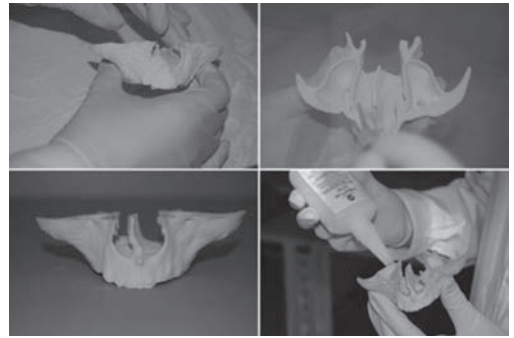


Figure 7. Cleaning, recycling and consolidation operations.

- Finishing that includes removing and recycling of excess material and model material consolidation.

Cleaning operations consist in the global remove, through compressed air action, of the non used dust to obtain an irregular but non dusty surface. After that, model surface consolidation is done by applying an epoxy resin or cyanoacrilate layer (Queijo & Rocha 2009).

3 CONCLUSIONS

3D replicas sections or global models are useful in diagnosing, planning and surgery simulation. The visualization and the possible manipulation, by patients, from 3D replica allow them to understand their pathologies nature, surgical proceedings that should be done by surgeon as well to reduce anxiety face to the surgery need. By other side it is possible to the surgeon to plan all the surgery procedures considering the chosen techniques and material that are dependant from the analyzed problem.

With multidisciplinary teams cooperation it is possible to build, in a short period of time, 3D

models that fulfill all the requirements to this applications.

In this case, by analyzing the global model and the five transversal sections—Figures 8 and 9, it is possible to, with a better accuracy, determine where the implants should be placed, considering the width of cortical bone available or, alternatively, determine where bone regeneration should be fomented to obtain a considerable cortical bone thickness that allows implants placement.

From that evaluation it became clear that the most suitable areas for implants placement would be the ones marked in the model and that can be seen in Figure 10.

Clinically and after evaluation of the bone condition in the 3D sections replica from the patient maxilla it has been decided to place 4 implants, trying to turn possible an acrylic prosthesis retention through attachments.

Implants will be placed in canine teeth area—two of them, while the other two will be placed in the second molar area, as can be observed in the figure. Due to the several patient bone tissues loss it should be used the techniques referred previously.



Figure 8. Manufactured and ready to use global 3D model.

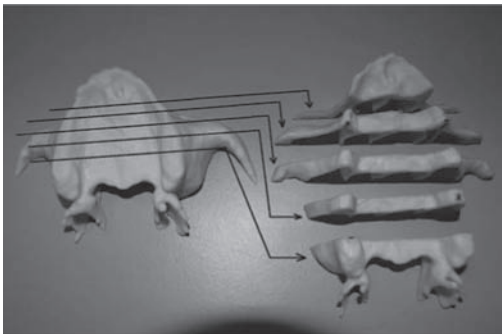


Figure 9. Global 3D model and corresponding transversal sections.

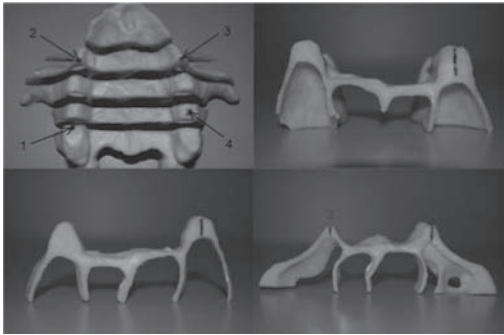


Figure 10. Preferred locations for bone regeneration and implants placement.

This way, in the posterior implants (Fig. 10—number 1 and 4) it will be used two Zimmer Tapered Screw-Vent® implants with 4,7 mm diameter and 11 mm long, by applying an osteotome bone condensation technique. Once the total height needed is about 5 mm the needed bone tissue will be gained by using osteotomes and applying Gen-oss®—Osteobiol® allogene bone.

In the anterior implants it will be performed a 2 steps surgery procedure. In first phase it will be opened a gingival side “window”. With proper instrumentation is slid the mucous membrane through which is applied Gen-oss®—Osteobiol® to perform bone regeneration. To close the open wound, a collagen membrane is applied to improve cicatrization and then, gingiva is sutured.

This last step will need a 6 months period of consolidation after which is needed a new evaluation to find out if the osseointegration was well performed and if no problems have arise since then.

In case of positive behavior confirmation it will be processed the surgery to apply the dental implants.

By showing these models to the patient he could understand the nature of his problems and have all the elements that allow him to choose if surgery should be considered or not. By not being an easy case, patient should always have the ability of choose if they want to proceed with the treatment and be aware of the risks involved.

Surgeon, become able to start planning surgery by estimating the work to be done and by doing one first evaluation to the place where he should intervene through this 3D replica.

In this line, and applied to the same subject, further work will consist in 3D modeling of medical devices (implants) and those will be inserted in the definitive positions assigned by medical staff allowing, this way, a more detailed surgical

planning with insertion points and angulations well determined as well with the definitive quantity of bone to be regenerated.

Another model will, then, be prototyped with all the areas aimed for intervention highlighted to better establish the definitive procedure.

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A numerical framework for wound healing at the bone-dental implant interface

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ABSTRACT: Osseointegration is the direct contact between bone and a dental implant. Its success depends upon the formation of new viable and functional tissues around the implant providing it with direct anchorage and stability. The aim of this work is to introduce a new mathematical framework based on reaction—diffusion equations describing the wound healing of the bone-dental implant interface prior to osseointegration. The set of equations here proposed is solved using the finite elements method. The results correspond to the distribution of the spatial-temporal patterns at the interface showing the ability of the model to qualitatively reproduce the biological features of blood clotting, osteoprogenitor cell migration, granulation tissue formation and bone matrix synthesis at the bone-dental implant interface including the surface topography of the implant. It is concluded that this framework can be used as a methodological basis for the formulation of a general model of the osseointegration in the bone-dental implant interface.

1 INTRODUCTION

A dental implant is a biomaterial device inserted in the jaw bone to replace a missing tooth achieving a firm, stable and long lasting connection with the surrounding bone in a process known as *osseointegration* (Schenk 2000, Albrektsson 2001). An adequate osseointegration is conditioned to the acceptance of the implant by the living tissues as well as to the formation of viable bone around the implant (Albrektsson 2001, Branemark 1983). This connection or bone-dental implant interface depends of biological and patient—related factors (Schenk 2000, Albrektsson 2001, Copper 1998), the implant design and surface (Matsuno 2001, Gapski 2003), the load distribution between bone and implant (Gapski 2003, Sikavitsas 2001), and the surgical procedure used for the implant placement (Fragiskos 2007, Branemark 1983).

During the wound healing and the new bone formation, several biological and biochemical events take place at the bone-dental implant interface. These events may be resumed into four stages (Figure 1) (Lang 2003, Aukhil 2000): 1) blood clot formation, 2) osteoprogenitor cell migration, 3) granular tissue formation, and 4) bone modeling. However, when a foreign device like a dental

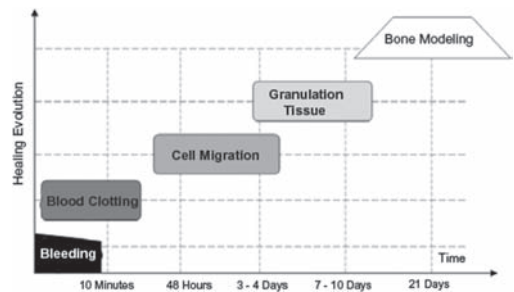


Figure 1. Time table of the healing evolution at the bone-dental implant interface (Ambard 2006, Aukhil 2000, Lang 2003).

implant is placed into the body, its surface interacts with the living tissues in a way that may change the profile of the events above described (Kasemo 2002, Davies 2003).

Thus, the success of the wound healing at the bone-dental implant interface depends of two additional phenomena: *osteinduction* and *osteochonduction* (Albrektsson 2001, Davies 2003). Osteoinduction is the recruitment of stem cells that are somehow stimulated to differentiate into bone—forming cells (Albrektsson 2001). If these

cells are capable of colonize the implant surface then this surface is said to be *osteoinductive*. Thus, osteoconduction means the bone formation over the implant surface (Albrektsson 2001, Davies 2003). This phenomenon essentially depends on the biocompatibility of the implant material and its surface characteristics (Huang 2005, Wennerberg 2003).

Despite of the great amount of knowledge coming from experimental models explaining the biological stages leading to new bone formation at the bone-dental implant interface (Aukhil 2000, Häkkinen 2000, Branemark, 1983), mathematical models and computational simulations provide additional information of the interaction between proteins, cells and tissues during the entire healing process (Moreo 2008, Ambard 2006). Here, we introduce a new mathematical framework describing the healing of the bone-dental interface as a sequential biological process mediated by the influence of the surface topography of the implant. Our framework describes by means of reaction–diffusion equations the aforementioned four biological stages of wound healing and the main ideas behind the osteoinduction and osteoconduction concepts. In the following section we briefly describe the biological process of bone formation at the bone-dental implant interface. Next, we introduce the mathematical formulation used and give details about its numerical implementation. Finally, we describe the results obtained, discuss some of the framework limitations and present our conclusion.

2 MATERIALS AND METHODS

2.1 Healing of the bone-dental implant interface

Although the biological process of healing at the bone-dental implant interface is highly complex, here we present a simplified version used as the theoretical support for our mathematical assumptions. A complete dissertation of the biology involved can be found in (Davies 2003, Aukhil 2000, Cochran 1999, Polimeni 2006), among many others.

The first event of the entire process is the surgical procedure used to place the dental implant (Figure 1). This is done by drilling a cavity in the jaw bone (Fragiskos 2007). After drilling, blood starts to flow and fills the placement cavity (Davies 2003). As blood flows, platelets start to aggregate and activate forming an initial plug that temporally detains the blood loss (Minors 2007). When activated, platelets release granules containing several chemical signals that control the migration of osteoprogenitor cells along the interface (Cochran 1999, Puleo 1999). The activation

of platelets is followed by a kinetic reaction that controls the conversion of *prothrombin*, a blood protein, into *thrombin*, which in turn reduces the blood protein *fibrinogen* into fibers of *fibrin* (Minors 2007, Gorkun 1997). The fibrin network finally created or *fibrin clot* replaces the platelets plug and provides support for the migration of osteoprogenitor bone cells towards the implant surface (Aukhil 2000, Davies 2003). At the time of the fibrin clot formation, the interaction of blood with the implant surface may increase the number of platelets activated with a subsequent increase in the concentration of the chemical signaling at the implant side (Kasemo 2002).

After these initial events, the presence of the chemical signaling induces the osteoprogenitor cells to start migrating from the bone surface over the fibrin network (Aukhil 2000, Cochran 1999). Almost concomitant with this migration, the necessity of vascular supply for the new tissues formation induces the prolongation of the blood vessels present at the bone side and the growth of a new vascular system along the interface (Sahni 2000). Together with this vascular formation, the production of a new connective tissue by the migrating cells gives place to the so-called *granulation tissue* that slowly replaces the fibrin network (Aukhil 2000, Häkkinen 2000).

Following migration, osteoprogenitor cells initiate their differentiation into osteoblasts to finally secrete new bone matrix. Since the osteoprogenitor cells arrive to the implant surface, the new bone formation or *osteogenesis* can be conducted in two ways, i.e., from the implant surface (*contact osteogenesis*) and from the bone side (*distant osteogenesis*) (Davies 2003). It has been demonstrated that rough implant surfaces induce platelet activation, osteoconduction and therefore contact osteogenesis (Davies 2007, Ellingsen 2006). Therefore, dental implants with rough surfaces tend to have more osteogenic cells over them and more concentration of chemical signals near them (Davies 2007). With the osteogenesis, the granulation tissue is replaced with a new bone matrix that with time is remodeled until the mechanical features of bone are restored and complete osseointegration is obtained (Branemark 1983, Schenk 1998, Sikavitsas 2001).

2.2 Mathematical model

For our approach to the biological problem we formulate the set of equations shown in Equations 1–8. Equation 1 and 2 correspond to the kinetic reaction between thrombin and fibrinogen, with T the thrombin concentration, P the prothrombin concentration, F the fibrinogen concentration and G the granules released by platelet activation. Parameters k_1 to k_6 control the production

and consumption terms. Equation 3 deals with the fibrin network formation where f is fibrin concentration, $f_{\max}$ is the peak fibrin concentration, w_f a fibrinogen threshold activation value, p a slope control value, α_f the fibrin conversion rate and β_f the blood quality factor.

Equation 4 describes the movement of osteoprogenitor cells concentration due to the gradient of the chemical signaling concentration described in Equation 5. Here Co is the concentration of cells, Qo is the chemical concentration, D_{Co} and D_{Qo} are the diffusion coefficients for cells and chemical, γ_{Co} is the cellular chemical sensitivity, new_{Co} and new_{Qo} are production factors for cells and chemical, $less_{Co}$, $less_{CxQ}$ and $less_{Qo}$ are consumption factors and $newQ(f, Co)$ a chemical-associated production term given as shown in Equation 6. At the implant surface ($X \in \Gamma$), $newQ(f, Co)$ is equal to a production factor new_{Qxl} multiplied by a surface roughness factor $factorS$, ranging from zero (smooth surface) to one (rough surface). By doing this, Equation 6 is able to describe the relation between roughness at the implant surface and chemical signaling activation. Furthermore, at the points of the interface where the levels of fibrin are over a certain threshold w_Q near its peak value ($X \in \Omega^D$) there is an increment in the chemical concentration given as new_{QxD} . This parameter represents the increment in the chemical concentration due to local accumulation of platelets over the fibrin network, their activation and the release of the chemical contents of their granules.

Equation 7 describes the granulation tissue formation along the fibrin network. Here TG is the granulation tissue density, new_{TG} is a production term associated to cell movement and fibrin formation, and $less_{TG}$ is a degradation term. Finally, Equation 8 is a transformation phase equation describing the formation of new bone matrix that replaces the granulation tissue, where B is the bone matrix density, α_B is a cell transformation factor and β_B is a chemical transformation factor. If only the chemical transformation factor is considered, which can be assumed to happen at the implant interface, Equation 8 represents contact osteogenesis. If both cell and chemical transforming factor are considered, i.e., both cells and chemical are present (which can be initially seen at the bone side of the implant), Equation 8 represents direct osteogenesis.

$$\frac{\partial T}{\partial t} = D_T \nabla^2 T + k_1 P - k_2 T - k_6 T F^2 \quad (1)$$

$$\frac{\partial F}{\partial t} = D_F \nabla^2 F + k_4 G + k_5 T + k_6 T F^2 - k_3 F \quad (2)$$

$$\frac{\partial f}{\partial t} = f_{\max} \frac{w_f^p}{F^p + w_f^p} (1 - \exp(-\alpha_f t - \beta_f f)) \quad (3)$$

$$\begin{aligned} \frac{\partial Co}{\partial t} = & \nabla \cdot (D_{Co} \nabla Co - \gamma_{Co} Co \nabla Qo) \\ & + new_{Co} Co Qo (1 - Co) - less_{Co} Co \end{aligned} \quad (4)$$

$$\begin{aligned} \frac{\partial Qo}{\partial t} = & \nabla \cdot (D_{Qo} \nabla Qo) + new(f, Co) \\ & + new_{Qo} Co - less_{QxC} Co Qo - less_{Qo} Qo \end{aligned} \quad (5)$$

$$newQ(f, Co) = \begin{cases} new_{Qxl} factorS, & X \in \Gamma' \\ new_{QxD} \frac{w_Q^p}{f^p + w_Q^p}, & X \in \Omega^D \end{cases} \quad (6)$$

$$\frac{\partial TG}{\partial t} = new_{TG} Co F - less_{TG} Co TG \quad (7)$$

$$B = 1 - \exp(-\alpha_B Co - \beta_B Qo) \quad (8)$$

3 NUMERICAL SIMULATIONS

3.1 Description of the simulation

The proposed model has been implemented using the finite elements method in a bidimensional domain of linear elements that reproduces a section of the interface between the border of bone and the screw-shaped surface of a dental implant (Figure 2). Geometry and dimensions of

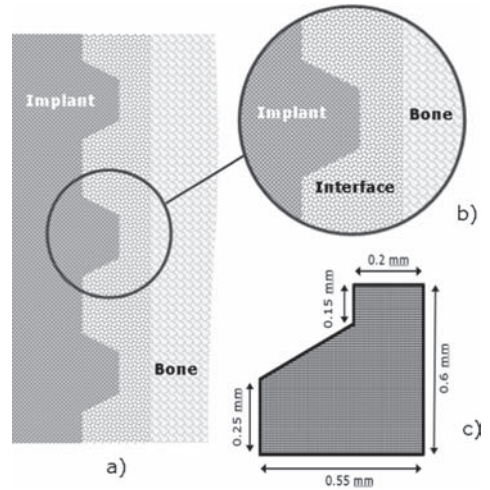


Figure 2. a) Sketch of a dental implant inserted in the jaw bone. b) Geometry of the interface formed between the bone edge and the implant surface. c) Diagram and dimensions of the interface section used in the simulation.

the implant outline were obtained from technical information provided by the dental implants manufacturer MIS Technologies Ltd. (Shlomi, Israel). The width of the interface is in concordance with experimental observations about the thickness of the protein layer created over the implant surface and the amount of necrotic tissue left by the insertion procedure (Kasemo 2002, Ellingsen 2006).

The mesh used is made of 18.728 quadrilateral bilinear elements and 18.417 nodes for Equations 1–3, 2.274 quadrilateral bilinear elements and 2.384 nodes for Equation 6, and 573 quadrilateral bilinear elements and 628 nodes for the rest of equations. The initial conditions for thrombin, fibrinogen and chemical concentration equations correspond to small perturbations of the steady state concentration obtained through numerical analysis of the reaction-diffusion equations (Vanegas 2009). For the remaining variables initial conditions are zero.

We have considered a flux condition at the bone side for the osteogenic cells concentration since cells are always present in the bone surrounding the implant (Davies 2003). Flux conditions for the rest variables were assumed equal to zero, therefore supposing the formation of an initial platelet plug that detains the blood loss in the injured area and the confining of the cells and tissues evolution to the analysis domain. The simulation time has been performed according to the time scheme shown in Figure 1. The computational solution was calculated using a PC with AMD Phenom X4 2.4 GHz processor, 8 GB RAM in a total running time of 2 hours.

3.2 Results

The model we present here simulates the blood clot formation, the osteoprogenitor cell migration, the granulation tissue formation and the new bone matrix synthesis at the bone-dental implant interface. Our approach lead us to include in the model the amount of platelets present in blood as a control for the fibrin network formation and the implant surface as a mediator for the chemical signaling of the osteoprogenitor cells. Figure 3 shows the fibrin

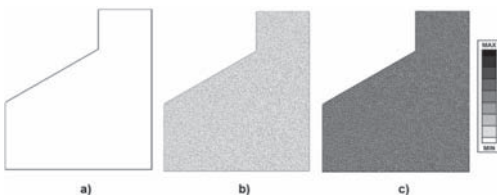


Figure 3. Evolution of the fibrin network formation. a) Initial condition at $t = t_0$. b) Fibrin network at $t = t_1$. c) Fibrin network at $t = t_2$. $t_0 < t_1 < t_2$.

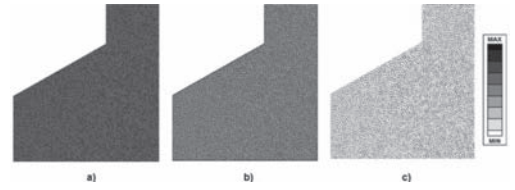


Figure 4. Steady-state of the fibrin network when the platelet concentration is modified. a) Thrombocytosis (1.000×10^9 platelets per liter). b) Normal level (300×10^9 platelets per liter). c) Thrombocytopenia (20×10^9 platelets per liter).

network formation as a consequence of the kinetic reaction between thrombin and fibrinogen. The dark zones correspond to fibrin fibers. In this case, the platelet concentration is supposed to be normal (around 300×10^9 cells per liter (Kaushansky 2005)). However, by changing the platelets concentration, the density of the fibrin network is modified. An increase in platelets concentration (600×10^9 – 700×10^9 cells per liter) causes *thrombocytosis*, a blood disorder that is usually asymptomatic but enhances the risk of thrombus formation, strokes and localized ischemia (Patel 2005).

In some cases of extreme thrombocytosis the number of platelets may be even above 1.000×10^9 cells per liter (Teffery 2007). In contrast, the diminishing in the platelets concentration (less than 150×10^9 cells per liter) or *thrombocytopenia* is a blood disorder characterized by the inadequate clot formation and the increased risk of bleeding (Patel 2005), which may be spontaneous if the number of platelets is less than 20×10^9 cells per liter (Mannucci 2000). Figure 4 shows the steady-state of the fibrin network for a platelet concentration of 1.000×10^9 cells per liter (Figure 4a) and 20×10^9 cells per liter (Figure 4c). Comparing these results with the steady-state obtained with a normal platelets concentration (Figure 4b) we can notice that the fibrin network in thrombocytosis is denser than in the normal level (more dark zones), situation that agrees with the risk of thrombus formation, as well as the less dense fibrin network of the thrombocytopenia case (more light zones) agrees with the risk of bleeding.

Once the fibrin network is created, the osteoprogenitor cells start migrating (Davies 2003, Aukhil 2000). As some platelets remain immerse in the fibrin network, their accumulation and posterior activation releases local amounts of chemical signals that induce cell movement. Figure 5 shows the evolution of cell migration and Figure 6 shows the evolution of the chemical signaling. The apparent distortion of the homogeneous concentration in Figure 5b and 6b is consequence of the platelet activation sites. Furthermore, during bleeding

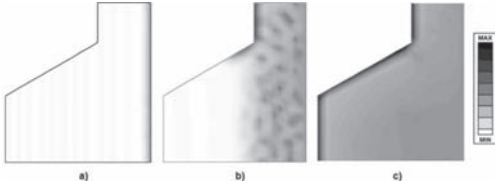


Figure 5. Evolution of the osteoprogenitor cells. a) Initial condition at $t = t_0$. b) Osteoprogenitor cells at $t = t_1$. c) Osteoprogenitor cells at $t = t_2$. $t_0 < t_1 < t_2$.

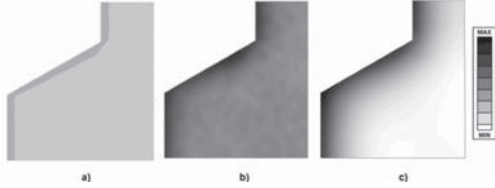


Figure 6. Evolution of the chemical signaling. a) Initial condition at $t = t_0$. b) Chemical signaling at $t = t_1$. c) Chemical signaling at $t = t_2$. $t_0 < t_1 < t_2$.

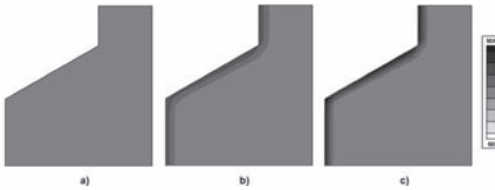


Figure 7. Steady-state of the osteoprogenitor cells at different types of implant surface roughness. a) Without roughness ($factorS = 0$). b) Medium roughness ($factorS = 0.5$). c) High roughness ($factorS = 1$).

platelets may reach the implant surface. If this surface induces platelet activation, the release of its granules increases the concentration of the chemical signal along the implant surface. Adopting the parameter $factorS$ as a surface roughness indicator, the behavior of the chemical concentration at the implant surface for different types of surface roughness can be achieved. Figure 7 shows the steady-state for the osteoprogenitor cells concentration for three different types of surface roughness. Here we can notice the ability of the model to simulate the osteoconductive property of the implant in terms of surface roughness since the greater the roughness the greater the osteoprogenitor cells concentration at the implant surface (Davies 2007).

As the osteoprogenitor cells migrate along the fibrin network, they differentiate and start producing a new connective tissue that replaces the fibrin scaffold (Häkkinen 2000, Aukhil 2000). At the same time, the demand of oxygen and nutrients

for the newly formed tissues causes the formation of a new vascular network inside the interface and along the fibrin fibers that initially support the vascular expansion (Lang 2003, Aukhil 2000). Figure 8 shows the formation of the new connective tissue as well as the restoration of the vascular supply. The appearance of the fibrin network pattern over the newly formed tissue represents the vascular formation along fibrin.

Once osteoprogenitor cells migrate through the bone-dental implant interface and the granular tissue is formed, the synthesis of new bone matrix begins. However, since the osteoprogenitor cells may reach the implant surface due to the activation of the chemical signaling over it, the new bone can be formed by means of either distance osteogenesis or contact osteogenesis. Figure 9 shows the bone formation by distant osteogenesis. As the front of

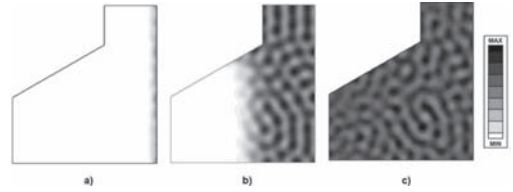


Figure 8. Evolution of the granulation tissue formation. a) Initial condition at $t = t_0$. b) Granulation tissue at $t = t_1$. c) Granulation tissue at $t = t_2$. $t_0 < t_1 < t_2$. The stripes pattern of b) and c) represent the formation of new vasculature along the fibrin network.

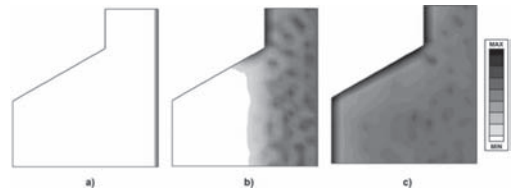


Figure 9. Evolution of distant osteogenesis. a) Initial condition at $t = t_0$. b) Distant osteogenesis at $t = t_1$. c) Distant osteogenesis at $t = t_2$. $t_0 < t_1 < t_2$. The dots in b) and c) represent the immersion of some osteoblasts into the bone matrix as osteocytes.

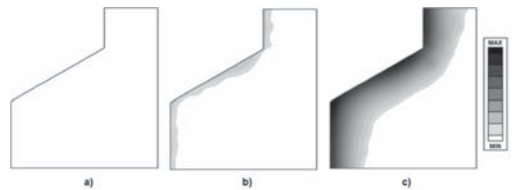


Figure 10. Evolution of contact osteogenesis. a) Initial condition at $t = t_0$. b) Contact osteogenesis at $t = t_1$. c) Contact osteogenesis at $t = t_2$. $t_0 < t_1 < t_2$.

new bone moves toward the implant surface, some of the producing bone cells become immerse into the newly bone matrix as osteocytes (Sikavitsas 2001). These osteocytes are represented in the simulation as the dot pattern shown in Figures 9b and 9c.

Figure 10 shows the time evolution of contact osteogenesis. Initially, there is no bone formation (Figure 10a). Since the implant surface induces chemical activation, the osteoprogenitor cells migration process lead to the cell colonization of the surface. As these cells differentiate and start forming new bone matrix, the front of new bone moves forward from the implant surface towards the bone border. As shown in Figures 10b and 10c, this contact osteogenesis bone front moves forward in a way that at the implant surface the amount of bone is higher than in the rest of the interface.

4 DISCUSSION

The aim of this work is to introduce a new mathematical framework for the numerical simulation of the wound healing in the bone-dental implant interface. The mathematical description is based under a simplified version of the biological events leading to the bone restoration at the interface, starting with the formation of the blood clot. Therefore, the results predict not only the formation of the fibrin network that compounds the blood clot but also the osteoprogenitor cells migration through the fibrin scaffold, the activation of the chemical signal leading cell migration, the implant surface interaction with this signaling, and the osteoconduction and osteogenesis phenomena. The main assumption of our approach is that all these biological events may be interpreted as sequential processes interacting in time and space. Hence, results are based upon the appearance of spatial-temporal patterns distributed along the bone-dental implant interface obeying to time restrictions.

Since the model is just an approximation to the realistic biological process of wound healing at the bone-dental implant interface, there are some limitations inherent to the chosen approach that should be highlighted. One of these limitations lies on the simplification made to the complex chain of biological events leading to bone formation at the interface. In our approach, we have not considered the degradation and cleaning processes prior to the vascular restoration that are started by the presence of macrophages and neutrophils during the clotting phase (Lang 2003, Davies 2003). Furthermore, we have simplified the several types of cells and chemical signals involved during the different stages of the healing processes (Puleo 1999, Cochran 1999, Aukhil 2000, Polimeni 2006)

as a simple osteoprogenitor cells contingent and a unique osteoprogenitor chemiotactic signal.

A second limitation relies on the fact that the mechanical stimuli modifies the behaviour of the biological phenomena of tissue healing and in consequence mediates in the adequate osseointegration of the bone-dental implant interface. Future work should be performed in order to include in the model the mechanical interactions in terms of cell adhesion, wound contraction and bone mechanosensing (Häkkinen 2000, Kasemo 2002, Sikavitsas 2001).

Regardless the limitations, the results here presented shown the ability of the model to reproduce important biological features of the wound healing at the bone-dental implant interface in concordance with experimental results (Branemark 1983, Lang 2003, Puleo 1999, Schenk 1998). According to this, we conclude that the mathematical framework described in this work is feasible to be considered as the methodological basis for the formulation a complete description of the wound healing and consequent osseointegration of the bone-dental implant interface.

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Evaluation of desinsertion strength of total prosthesis with an intra-oral transducer

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ABSTRACT: The use of adhesives as an auxiliary way of prosthesis' retention has increased over the years. This can be explained by the functional implications that arise from poor retention/stability of total removable dentures and by the improvement of adhesive proprieties of this kind of materials. The main aims of this study were: to test the use of an intra-oral transducer in evaluation of the strength of desinsertion of total maxillary prosthesis; to evaluate the effect of adhesives on retention of these prostheses and to compare the results of different adhesives tested. The evaluation was done by quantifying the strength of the prosthesis desinsertion when subjected to vertical tensile tests, using, for this purpose, an intra-oral resistive transducer with 4 extensometers (120 Ω), forming a Wheatstone's bridge. All adhesives improved retention of complete upper dentures. The intra-oral transducer used has proved to be an appropriate instrument for the aims of this study.

1 INTRODUCTION

Improving denture retention and stability has been a challenge in prosthodontics over the years (Psillakis 2004). The first prosthetic adhesive appears in the eighteenth century but the first patent by American Dental Association (ADA) only appears in 1913. The purpose of the use of denture adhesives can be described as to subjectively benefit denture-wearers with improved stability, retention and comfort of their dentures, and with improved incisal force, masticatory ability, and confidence (de Baat 2007, Ozcan 2005). Despite the more or less restraining attitude of dentists towards denture adhesives, it has been shown that a substantial proportion of denture wearers had tried denture adhesives in the past or is regular users currently (de Baat 2007, Coates 2000).

2 AIMS

The main aims of the authors of this research were: to test the use of an intra-oral transducer in the evaluation of the strength of desinsertion of total maxillary prosthesis; to evaluate the effect of adhesives on retention of these prostheses and to compare the results of different adhesives tested.

3 MATERIAL AND METHODS

The retention of maxillary complete denture was evaluated, initially without adhesive and then with each of the 5 adhesive studied (1- Protefix cream, 2- Corega cream, 3- Corega ultra powder, 4- Protefix powder and 5- Corega strips—Figure 1), in a sample of 26 patients from FMDUP. In the end of the tests another evaluation without adhesive was done to be sure that no adhesive rest between the different tests.

The evaluation was done by quantifying the strength of the prosthesis desinsertion when subjected to vertical tensile tests, using, for this



Figure 1. The five adhesives tested.

purpose, an intra-oral resistive transducer with 4 extensometers (120 Ω), forming a Wheatstone's bridge. In the center of the prosthesis, finding thus determination of the center of one triangle defined by three points (tuberosities and interincisal papilla) we put a rivet to receive a screw from the transducer—Figure 2.

A data acquisition plaque and the LabView software enabled the reading of values that subsequently were treated statistically in the SPSS program—Figure 3.

4 RESULTS

The intra-oral transducer used has proved to be an appropriate instrument for the aims of this study. All adhesives improved retention of complete upper dentures, although Corega strips do not do

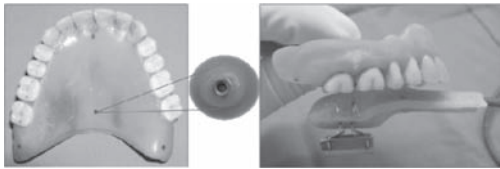


Figure 2. Determination of the place of the traction and connection to the intra-oral transducer.



Figure 3. Data acquisition card and reading the values obtained.

it in a statistically significant way. The adhesives Corega cream, Corega powder and Protefix powder showed very similar retention values. The values of initial and final tests without adhesive were equivalent (Table 1).

5 DISCUSSION AND CONCLUSIONS

Since the existence of removable prosthesis that dentists seek to improve retention and stability. The evaluation of these parameters has been made in various ways over the time. Once the retention is that quality inherent in the dental prosthesis acting to resist the forces of dislodgment along the path of placement and the stability the quality of a removable dental prosthesis to be firm, steady, or constant, to resist displacement by functional horizontal or rotational stresses (GTP-8 2005), easily we understand that studies with the utilization of gnathometers (Ozcan 2005), kinesiography (Psillakis 2004, Grasso 2000, Grasso 1994) and infrared (Hasegawa 2003) evaluate the stability in detriment do retention. We understand that these two parameters are very difficult to dissociate in clinic tests, but with an intra-oral resistive transducer like the one we used is possible analyzed only the retention of total prosthesis.

Although adhesives are poorly recommended by the oral care professionals by an attitude of criticism, indifference and/or ignorance, when used correctly, they can be an advantage in retention and stability of removable prostheses in general and total prostheses in particular (Psillakis 2004, Coates 2000).

The improvement promoted by adhesives in satisfaction and retention is more pronounced in the maxillary than in mandibular denture (de Baat 2007). That's why mandibular overdentures and implants first emerged (Psillakis 2004). With

Table 1. Pairwise comparisons.

(I) Factor 1	(J) Factor 2	Mean difference (I-J)	Std. error	Sig. ^(a)	95% Confidence interval for difference	
					Lower bound	Upper bound
Without adhesive initial	Protefix cream	-540.091(*)	146.196	.027	-1041.866	-38.317
	Corega cream	-829.222(*)	158.329	.001	-1372.639	-285.805
	Protefix powder	-762.235(*)	169.082	.004	-1342.560	-181.910
	Corega powder	-968.591(*)	119.645	.000	-1379.238	-557.944
	Corega strips	-521.957	188.915	.238	-1170.350	126.436
	Without adhes. final	-49.070	139.087	1.000	-526.446	428.306

Based on estimated marginal means.

*The mean difference is significant at .05 level.

^aAdjustment for multiple comparisons: Bonferroni.

this kind of intra-oral transducer we could not evaluated the retention of mandibular prosthesis but to the future we hope to developed a new transducer smaller and easily to applied that allow us to do that study.

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Analysis of the orthodontic wire behavior through the computational numerical simulation

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ABSTRACT: Load determination for orthodontic wires by means of experimental testing has received, in recent years, the attention of many researchers. Using the mentioned tests it is possible to measure the loads of a system equivalent to the constituted in the oral cavity and thus, indirectly, to get information of the orthodontic movements. In this work, numerical simulations of models of experimental orthodontic wire tests are presented in order to get important information of the wires. The present study is based on mathematical models used to simulate tests of deflection of orthodontic wires, through the Finite Element Method. The validation of the models was carried out through comparison with the results from the literature. The results demonstrate the capacity of the models to represent the real physical problem and the potential of this tool for the study of the behavior of orthodontic wires.

1 INTRODUCTION

One of the main objectives of the Orthodontics is the correction of the positions of teeth by means of the application of controlled efforts on these and, unquestionably, the most used form to this is the orthodontic wire under activation. There is a necessity, therefore, to know the behavior of orthodontic wires, to make more accurate the movement of teeth (Sernetz, 2005).

The basic principle of application of efforts on teeth is the rabbit of these, under deflection in to the brackets. The efforts magnitude depends mainly of the mechanical properties of the wires, that can be express for the relation load for millimeter (N/mm), thus representing the rigidity of each used wire (Asgharnia e Brantley, 1986).

Thus in the dependence of a rigidity of the wire, influenced directly for the constituent alloy and the transversal section of this, different efforts, for one same deflection, are generated by the orthodontic devices. This makes with that it is basic the knowledge of the elastic properties of each one of these wires so that the professional selects the wire most adequate (Quintão, 2000).

Most of the studies on mechanical behavior of wires used in Orthodontics, is carried out by means of laboratorial experiments of deflection. However,

other methods also can be used to represent the equivalent systems to these in orthodontic treatments and thus to supply useful information on the mechanical behavior of these (Gurgel, Ramos e Kerr, 2001).

Although the versatility and of the great advantages offered for the Finite Element Method (FEM) for assays of this nature, does not exist a great number in the literature of research that simulates such types of assays for its. Through this method therefore, as well as the experimental studies, in the simulations of deflection of wires, it can be simulated then what it happens in the clinical routine during the appliance activations, representing the most varied combinations of alloys and dimensions of wires (Gravina et al, 2004).

This becomes the MEF a reasonable alternative, without increases the costs or difficulties of procedures, for these ends. It can be considered a tool in the aid of the agreement of the results of the some studies made on the mechanical behavior of orthodontic wires, for the reproduction of the same ones. (Ferreira, 2001).

Based in the importance of the mechanical behavior of orthodontic wires, over all in the initial phases of treatment (alignment and levelling), and in the resources offered for the Finite Element Method, with the present work it is intended to carry out

computational numerical simulations of orthodontic wire under activations being compared the results with the previous studies published and analytical solutions of form to validate the models considered.

2 OBJECTIVES

The specific objectives of this work are:

To consider a bidimensional model of finite elements for the study of the mechanical orthodontic wire behavior, simulating the experiments deflection in three points;

To simulate orthodontic titanium molybdenum wire under deflection in three points of transversal sections of $0.41 \text{ mm} \times 0.56 \text{ mm}$, $0.43 \text{ mm} \times 0.64 \text{ mm}$, $0.53 \text{ mm} \times 0.64 \text{ mm}$, $0.46 \text{ mm} \times 0.46 \text{ mm}$, $0.53 \text{ mm} \times 0.53 \text{ mm}$ and to compare the results with previously found by Johnson 2003 and with analytical solutions, in order to validate of the considered model;

To make a sensitivity analysis of the model in result of the variation of the length of the wire segment between the supports.

3 METHODOLOGY

To reach the considered objective had been developed, in this research, activities related to the following stages: (1) Model – Johnson 2003; (2) Analytical solutions; (3) Comparison of the results and (4) Model – ISO 15841.

3.1 Model – Johnson 2003

The geometric characteristics of the model initially considered in this study had been gotten of the descriptions of the experimental procedures of an orthodontic study published by Earl Johnson in the Angle Orthodontist journal, in 2003. Figure 1 shows the device used in the experiments of Johnson.



Figure 1. Device for deflection in three orthodontic wire points used in the experiments of Johnson, 2003. The center of the wire (distant 6 mm of each lateral support) was submitted to a deflection. The extremities of wires beyond the supports had not been fixed (Johnson, 2003).

The author tested Titanium Molibdenum alloy (TMA) orthodontic wires, rectangular and square transversal sections of $0.41 \text{ mm} \times 0.56 \text{ mm}$, $0.43 \text{ mm} \times 0.64 \text{ mm}$, $0.53 \text{ mm} \times 0.64 \text{ mm}$, $0.46 \text{ mm} \times 0.46 \text{ mm}$, $0.53 \text{ mm} \times 0.53 \text{ mm}$.

These values associates to the dimensions of the device of presented assay had been adopted for the bidimensional model. The wire with restrictions of vertical translation in the points of the supports and a displacement of 1 mm to the center of the supported segment was used.

The reaction of the supports had been registered and represented thus the force for the wire 1 mm of activation.

3.2 Analytical solutions

Supported beam was submitted to a central displacement of 1 mm was used to calculate. Elasticity Modulus of TMA, the dimensions of the transversal section of used wires, and the Moment of Inertia for each submitted section had been use.

3.3 Comparison of the results

In this stage the comparison of the results obtained in numerical simulations was carried out, with analytical solutions and with the experimental studies of Johnson 2003.

3.4 Model – ISO 15841

In this phase of the research, for the final modeling and simulations, have been used the specifications contained in Norm ISO – 15841 for elastic wires (Type 1). The distance of 10 mm between brackets of neighboring teeth is represented. This was practically the only difference of the model based on the work of Johnson 2003.

The flexural rigidity was determined (N/mm) in the ocluso-gingival direction as it praises the norm. The Figure 2 shows the representative project of the orthodontic wire tests ISO 15841.

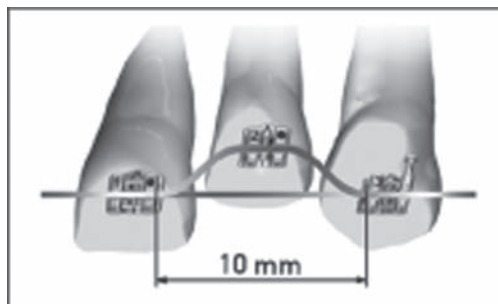


Figure 2. Representative project of the orthodontic wire tests according to ISO 15841. (Sernetz, 2005).

4 RESULTS AND DISCUSSION

The results of the simulations of this study had been vertical forces generated by orthodontic Titanium Molybdenum alloy (TMA) wires under displacement of 1 mm of activation in three points deflection. Figure 3 shows to the model considered with the displacement of 1 mm in the central region and the two lateral supports. In Figure 3 the model deformed after can also be observed.

In this work basic however fundamental aspects in the analyses of the mechanical behavior of orthodontic wires, will be presented and argued. In example, the dimensions of the wire, and the distance of the segment between the supports and as its influence in the results.

In the results of the numerical simulations of the present study, as already he was waited, the values increased when to increase the Moment Inertia of the Section (MIs) values, both the Simulations J and Simulations ISO (simulations followed the characteristics of the experiments of Johnson (2003) and followed the specifications of the ISO-15841 (2006), respectively.

The Table 1 shows the results of vertical force in Newton (N). The result relative to the displacements of 1 mm for all the carried out simulations.

In Graph 1 the results of the simulations are plotted in function of the Moments of Inertia of the

sections (MIs). In this graph (1) the preview relation of proportionality can easily be observed among the values of MIs for the five analyzed dimensions and the forces of the simulations. In increasing sequence: (MI 1) 0.41 mm × 0.56 mm; (MI 2) 0.46 mm × 0.46 mm; (MI 3) 0.43 mm × 0.64 mm; (MI 4) 0.53 mm × 0.53 mm e (MI 5) 0.53 mm × 0.53 mm (MI 4).

When the simulations results had been compared with the results of analytical solutions had revealed sufficiently similar, what it contributed for the validation of the model. In Graph 2 the relation of the values for simulations of section wires of 0.41 mm × 0.56 mm can be observed. Both the Simulations J and Simulations ISO had demonstrated values well next to the values of analytical values. What one more time comes to corroborate to validate them as tools of study of the mechanical behavior of orthodontic wires.

It can be observed however that although to coincidence of the results of the analytical results with the Johnson 2003 and ISO 15841 results, these differ sufficiently between itself.

In the simulations of Johnson 2003 one aspect was verified that it deserves to be salient, the distance of kept segment of the one of wire between supports in the experiments. In the present work was made a sensitivity analysis of the model considered this parameter.

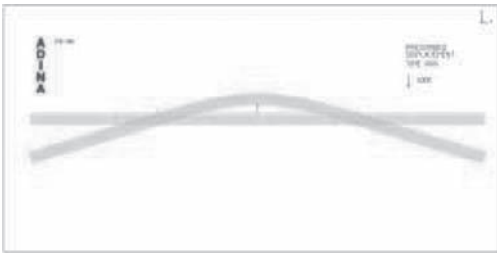
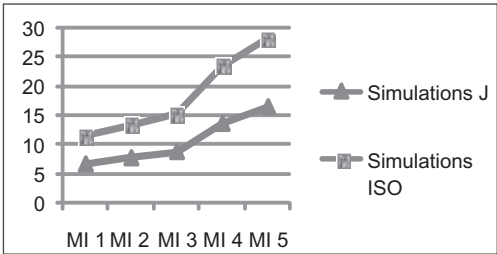


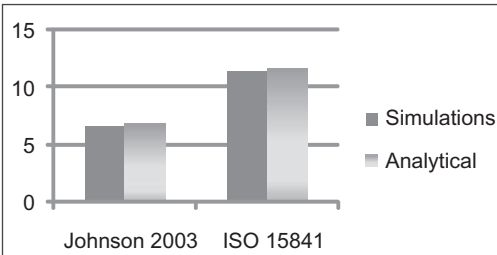
Figure 3. Considered bidimensional model. It can be observed the representation of the displacement to the center and the supports in the laterals. An overlapping sight of the model after load also can be verified. (ADINA® 8.5).

Table 1. Results of the simulations.

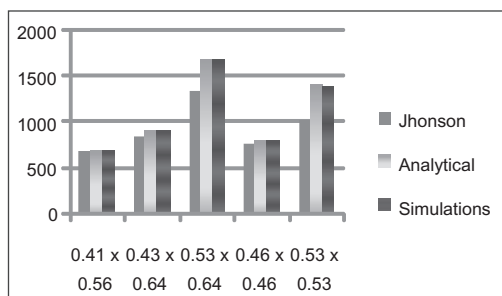
Dimension of wire	Simulations J		Simulations ISO	
	N	Grams	N	Grams
0.41 × 0.56	6.67	681	11.52	1117
0.43 × 0.64	8.80	898	15.18	1548
0.53 × 0.64	16.45	1678	28.34	2891
0.46 × 0.46	7.74	789	13.35	1362
0.53 × 0.53	13.62	1389	23.47	2394



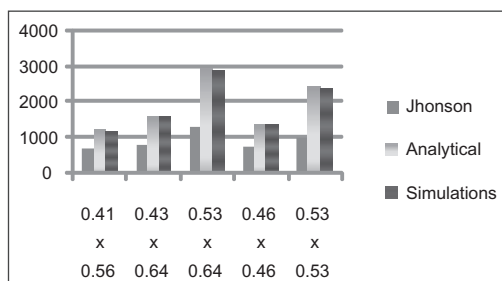
Graph 1. Results of the Simulations of this study in function of the value of the Moment of Inertia of the section (MI).



Graph 2. Results of in the Simulations J, Simulations ISO and Analytical results for wires of 0.41 × 0.56 mm.



Graph 3. Results of the Simulations J, Simulations ISO and theoretical results to 12 mm.



Graph 4. Results of the Simulations J, Simulations ISO and theoretical results to 10 mm.

The distance between the supports during the assays of Johnson (2003) was of 12 mm in the place of 10 mm, praised for ISO 15841, and although this difference to be small, considered for the model a reasonable reason for a sensitivity analysis of the model front the variations of this parameter.

In the comparison of the simulation results varying only this parameter (distance between the supports), was verified that differences of 41.9% for wires of dimensions 0.53×0.64 mm and 42.2% for wires of dimensions 0.41×0.56 mm was found, what means to say that the in front of variation of such parameter, this model showed sufficiently sensible. This occurred as already waited, and this occurred mainly to the dimensions very reduced of the model front which 2 mm of alteration already sufficiently modifies the results.

This research showed what the on mechanical orthodontic wire behavior studies must rigorously follow the specifications of the Norms, as of ISO 15841. Of the opposite, similar wires in different situations as for example in the distance between the supports of 10 mm, 12 mm, wires of same dimension and alloy can generate resulted

well different and in place of assisting, it still to complicate the election of wires for the professional of the Orthodontics.

5 CONCLUSIONS

In view of the results of this work one concludes that:

The bidimensional model in finite elements considered in this study assists in the study of the mechanical orthodontic wire behavior, simulating the experiments deflection in three points;

The simulations of the TMA wires assays of the diverse rectangular and square transversal sections was compared with analytical and experiments of Johnson 2003, and showed a similar values, therefore this served of definitive form for the validation of the considered model;

The results of sensitivity analysis of the of the model in front of the variation of the length of the wire segment was carried out and was verified that an modification of 2 mm can be generated forces very different. Therefore must rigorously to follow the Norms during any experimental or numerical assay.

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Characterization of polymerization reaction of self-curing dental cements using fibre optic sensors

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ABSTRACT: The aim of this study was to monitor the shrinkage during the polymerization reaction of self-curing dental cements, using optical fibre sensors. With the present technology was possible to measure, in real time, shrinkage and temperature variations during chemical reaction. The knowledge of such properties can be useful in the prevention of possible leakage and failure in dental restoration procedures.

1 INTRODUCTION

In the last years, dental composite cements have been commonly used as restorative dental materials and also applied in endodontic, orthodontic and periodontic purposes.

The knowledge of their properties and proper manipulation is essential to obtain the best results in clinical dentistry. The bad choice of the material to apply can lead to microcracks at the tooth composite interface or leakage and failure of the restoration.

Initially it was considered three main categories of cements, namely zinc phosphate, zinc oxide eugenol and silicate. In 1950's appear a new class called resin cements. This kind of cements is constituted by two different materials, one of which containing benzoyl peroxide initiator and the other one, a tertiary amine activator. When the two materials are mixed the amine reacts with the benzoyl peroxide to form free radicals and the polymerization is initiated. Cements based on glass ionomer are other category that emerged in 1970's. They were formulated by adding polyacrylic acid as the liquid component to finely ground silicate powder. The setting reaction of these cements is an acid-base reaction between the acidic polyelectrolyte and the alumina-silicate glass. In practical terms, this means that a powder is mixed with a liquid to

generate a mixture which then sets and hardens through a setting reaction (Phillips 1991).

In this work, the changes in the linear shrinkage and variation of temperature that occur during the polymerization reaction of two different categories of self-curing dental cements were monitored, in real time, using Fibre Bragg Grating (FBG) sensors. This technology has already shown promising results in the study of dental materials (Anttila et al. 2008); (Arenas et al. 2007); (Milczewski et al. 2006).

2 EXPERIMENTAL PROCEDURE

2.1 Materials

With the intention of comparing results, two different categories of dental cements were studied, namely resin cement and glass ionomer cement. It was tested Cement-post manufactured and commercialized in Brazil by Angelus® and Ketac cem commercialized in Portugal, but manufactured in Germany by 3M ESPE®, respectively, as example of the two categories mentioned. Both cements are chemically activated, being the former supplied as two separately pastes, namely catalyst paste and universal paste and the other one is available in the form of powder and liquid.

Among several applications, these cements are used in the cementation of inlays, onlays, crowns and pins.

2.2 Fibre Bragg Grating sensors

A Fibre Bragg Grating consists in a periodic perturbation of the refractive index along the fiber length, which is formed by exposure of the core to an intense optical interference pattern of ultraviolet radiation.

The principle of operation of an FBG sensor is based in the Bragg condition that is given by the following expression:

$$\lambda_B = 2 n_{\text{eff}} \Lambda \quad (1)$$

where λ_B is the reflected Bragg wavelength, n_{eff} is the effective index of the core of the fibre and Λ is the refractive index modulation period.

When a FBG is illuminated by a broadband source, only the Bragg wavelength that satisfies Equation 1 is preferentially reflected, being all the others transmitted. If the grating is subjected to mechanical perturbation or thermal variations, the reflected Bragg wavelength is modified, so it is possible to employ these elements as sensors.

In our case, two FBGs sensors were used, being the gratings inscribed into photosensitive single mode optical fibre, by illuminating it with ultraviolet CW laser irradiating at 244 nm, using an automated phase-mask interferometer system (Nogueira et al. 2002).

One of these sensors, called temperature sensor was placed inside a double needle (Figs. 1, 2), in order to be protected mechanically and to be only sensitive to variations of this parameter. The other sensor consisted in an FBG in direct contact with the cement and is sensitive to temperature and strain variations. This last sensor was placed inside a silicon mold that contains a hole in the bottom, through which the fibre is pulled through. The fibre was bonded to the support and pre-tensioned. The temperature sensor was also placed inside the mold.



Figure 1. Temperature sensor.

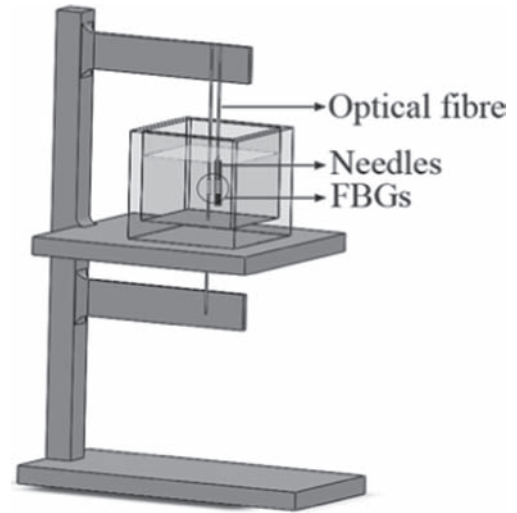


Figure 2. Schematic representation of the experimental setup.

2.3 Measurement of temperature and strain variations

The cements were prepared according with the manufactures's recommendations that, in the case of Cement-post cement corresponds to mixture equal amounts of base and catalysts pastes and spatulate during 10 seconds. Concerning to Ketac cem cement, the powder-to-liquid mixing ratio was one level spoonful of powder for two drops of liquid, being necessary to spatulate during 1 min.

After the mixtures preparation, these were flowed into the silicon mold and data acquisition was made at 1 sample/second, during 150 min, simultaneously for the two sensors.

The measurements were carried out for several samples of each cement. The final results presented in this work correspond to the average of the total amount of measurements.

3 RESULTS AND DISCUSSION

The evolution of strain and temperature, during the polymerization reaction of the two different dental cements studied, is presented in Figures 3, 4.

The strain curve was obtained by subtracting the effect of temperature, obtained with the temperature sensor, in the measurements accomplished by the other sensor, which is sensitive to both temperature and strain variations.

The obtained results show an initial increase of temperature that reaches 18.5°C and 14.3°C for

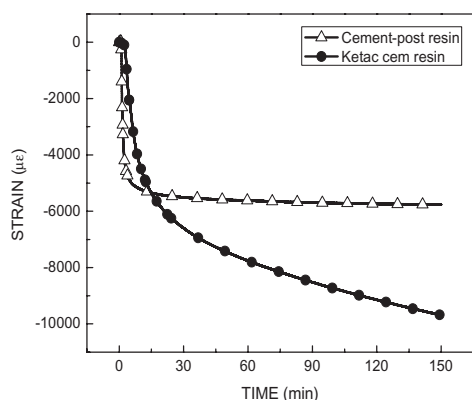


Figure 3. Evolution of strain during the polymerization reaction of the two dental cements.

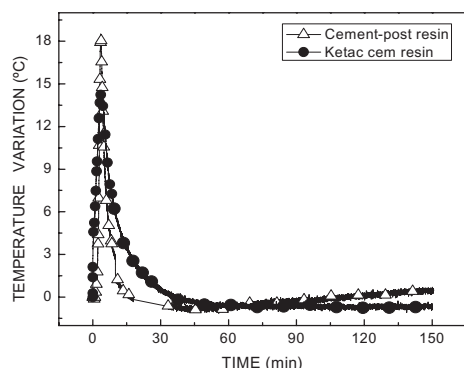


Figure 4. Evolution of temperature during the polymerization reaction of the two dental cements.

Cement-post and Ketac cem cements, respectively, after 4 minutes of starting setting reaction. The polymerization reaction of these cements is then exothermic. During this temperature increase, an accentuated shrinkage of the material is also observed, which continues even after temperature reaches its maximum and starts to decrease back to room temperature. After 15 min of experiment, both cements present about of 5350 $\mu\epsilon$ of shrinkage. In the case of Cement-post cement this strain value maintain practically unaffected to the end of the experiment. Concerning to the other cement, although it has reached about -10,182 $\mu\epsilon$ at the end of 150 min of setting time, according with the graph of Figure 3, the tendency is to continue to shrink. At the end of the experiment, the Ketac cem cement presented almost the double value of strain than Cement-post. The shrinkage is more effective in the first cement and by this fact, it can lead to worse performance for dental applications of this material.

4 CONCLUSION

The application of Fibre Bragg Grating sensors to monitor, in real time, the polymerization reaction of chemically activated dental cements were presented.

The two cements studied reached different values of temperature and shrinkage so, it is important to know the performance of the materials before to be used, in order to obtain the best results.

The technology applied in this work promises to be a simple method to determine simultaneously shrinkage and temperature variations of dental cement, being suitable to be applied in the study of other dental materials. The supplied information can be used to optimize their properties and application methods.

ACKNOWLEDGEMENTS

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Clinical evaluation of implant retained overdentures: Biological complications

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ABSTRACT: Implant retained overdentures is an effective method of treatment for edentulous patients when unstable prosthesis and residual ridge resorption are present. Usual complications mostly consist of screw loosening, gingival hyperplasia, and needs for replacement of clips, denture relining and readjustment of occlusion and articulation. This study aims to report the clinical biological complications and to suggest possible relations with functional, mechanical and technical aspects. Twenty-one files of patients treated with removable acrylic implant-retained prosthesis using clip-bar attachment system were evaluated. The incidence per patient varies along the prosthetic rehabilitation life-time. In the first year (mean/patient), the loss of adaptation (0.29) and soft-tissue damages (0.10) were the main problems. In further time the denture related stomatitis earns a more important incidence. The bone loss increased along the years. One implant was lost. The biological complications are a common important event, but many times minimized by the professionals. Close follow-up is inevitable.

1 INTRODUCTION

The use of overdentures retained by implants, often in reduced number, 2 to 4 implants, is an effective method of treatment, safe and with good predictability to edentulous patients, especially when unstable removable prosthesis and mild to severe residual ridge resorption are present (Assunção 2008). It offers a solution with a high longevity, great acceptance and satisfaction, preserving the residual alveolar ridge, improving retention, stability and chewing efficiency (Assunção 2008).

The “Splinting bar-clip” system, although expensive and technically demanding, is one of the most used and reliable, avoiding difficult and unpredictable surgical interventions (Karabuda 2008).

Despite complications are usual, they are frequently not serious, consisting of screw loosening, gingival hyperplasia, and needs for replacement of clips, denture relining and readjustment of occlusion and articulation (Geertman 1996).

2 AIM

The aims of this study are to report the biological complications that arise in the clinical practice and to suggest possible relations with mechanical and

technical aspects of implant retained removable prosthesis and muscular function.

3 MATERIAL AND METHODS

There were evaluated, retrospectively, the clinical and panoramic radiology files of 21 rehabilitations, treated in private practice in a Dental Office in Oporto, Portugal, with removable acrylic implant-retained prosthesis using clip-bar attachment system, supported by different number of implants with different conformations of the bar, height and thickness.

The selection criteria were to include all patients treated this way, who had appointments between October 1st, 2008 and February 1st, 2009.

The antegonial notch was radiologically evaluated as a sign for muscular hyperactivity (coded 0 for flat and 1 for prominent), as well as the bone loss around implants (coded 0 for absent of vertical signs of bone loss, 1 for reduction inferior to $\frac{1}{3}$ of implant length, 2 for $\frac{1}{3}$ to $\frac{1}{2}$ and coded 3 when superior to $\frac{1}{2}$ of implant length). The type of opposing dentition, new prosthesis and the biological and mechanical complications that these rehabilitations presented over the time were pointed, and evaluated clinically or radiologically.

Statistical treatment was performed using Microsoft Office Excel 2003 and conventional statistical formulas.

4 RESULTS

The main biological complications that were found are bone loss, loss of adaptation of prosthetic components, loss of implants, soft-tissue lesions and denture related stomatitis. The incidence per patient of these different complications varies along the prosthetic rehabilitation life-time.

In the first year after the prosthesis was set, the loss of adaptation (0.29) and soft-tissue damages (0.10) were the main problems. In further time the denture related stomatitis earns a more important incidence (Figure 1). Only three patients had been diagnosed with denture related stomatitis, and in only one case was observed at least a once a year recurrence for last five years. The bone loss has progressively increased along the years (Figure 2) still, the bone loss until half of length of the implants is a frequent event. Logistic regression model was not conclusive to establish any relation between antegonial notch (sign for muscular hyperactivity), biological complications and bone loss. An only one in eighty-one implant was lost. This implant has been lost after 9 years of function.

5 DISCUSSION

According to the literature, the majorities of the complications described are more related with

the prosthetic components than related with the abutments and support tissues. Biological complications and their relation to mechanical aspects are the less considered in the overdenture-related literature available (Berglundh 2002). Opening the speculative idea that biological implication over the arches and the supporting tissue around the implants, particularly in extreme circumstances, or in case of deficient surgical planning/execution, is being ignored. In any circumstance, the biomechanics is a critical factor in all stages, from the planning and treatment to the follow up.

The excess of functional loads generates stresses that are dissipated by the retention system and consequently by the implants and supporting tissue. The intensity and amplitude of bone resorption is determined by the transmission and distribution mechanism of each retention system (Assunção 2008).

Even having in mind that the type of soft-tissue complications is varying through studies, the results that are being provided in this study fit the general expectations but not without presenting some discrepancies with common literature results (Berglundh 2002). This 2002 literature-review by Berglundh, revealed in their meta-analysis the following weight mean for biological complications: a) 5,86% of implants lost during function (being the most frequent, the ones placed in the maxilla, and twice as high in overdentures than in fixed reconstructions); a mean of 0.27 soft-tissues complication a patient in 5 year period; Peri-implantitis of 0.66% and 4,76% of the implants with a bone loss superior to 2,5 mm (Berglundh 2002).

The main biological complications we reported were bone loss, loss of adaptation of prosthetic devices, loss of implants, soft-tissue lesions and denture related stomatitis. In the first year the biological complications were especially related to the biological adaptation of the prosthesis, and later to prosthetic stomatitis (Figure 1). It is important to mention that hyperplasia is one a common finding in patients with a bar supported prosthesis (Berglundh 2002, den Dunnen 1997). As expected the bone loss has progressively increased along the years (Figure 2). It was not possible to obtain any correlation (logit model) involving antegonial notch (as marker for muscular hyperactivity), bone loss and biological complications. The first year is critical, since most of the adaptations and the larger number of problems took place (Karabuda 2008, Berglundh 2002, Geertman 1996). In any way, as concluded in the study from den Dunnen et al., the results from the present evaluation confirm the need for close routine follow-up services of hygiene care, adjustments, and treatment of complications (den Dunnen 1997).

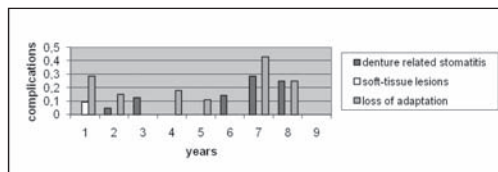


Figure 1. Mean values of biological complications.

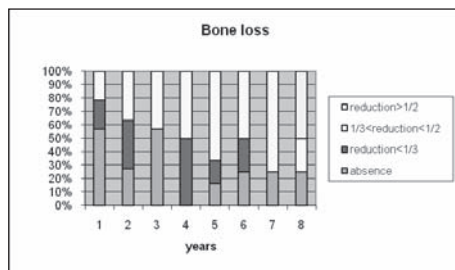


Figure 2. Evidence of vertical signs of bone loss along the years in proportion to implant length.

Despite the fact that different methodology is being used, the results of this evaluation point to clinicians' lack of attention on soft tissue (den Dunnen 1997, Berglundh 2002, Assad 2004).

In the planning, it is crucial to attend the biomechanical concepts, including face type, mastication type and strength, the axis and distribution of forces, the type of occlusion, the construction materials of infra and suprastructures, as well as the types of plug-in, the number and distribution of implants, the antagonist tooth quality and, very important, the support tissues and the adaptation to oral soft tissues, in order to be possible to ensure a predictable and sustainable rehabilitation.

6 CONCLUSION

It is consensually accepted that implant retained or supported overdentures require frequent maintenance and careful hygiene, being the most important problems disclosed in the first year of use. Thus the need for close follow-up is not only inevitable because of frequency of complications as it is intrinsic and should be accepted for both clinicians and treated patients as normal event in the use of this valuable kind of rehabilitation. Most biological complications have inherent mechanical aspects and complications.

The involvement of oral tissues in implant-retained removable rehabilitations lacks more studies, particularly in what respects to biomechanics, planning and the longevity of the works. The implications over biologic tissues, especially in the classic cases of the use of two implant abutments, with

mucosa support, or the less soft-tissue demanding, "implant-supported", situation with four implants, still unclear.

The biological complications are a common, important event, but many times minimized by the professionals, who tend to conceptualize this kind of rehabilitations as a fixed one or tend to be more focused in the prosthetic failures or malfunction, neglecting the biological relevance for biomechanical success and durability of the implant retained overdentures.

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Case study of a temporomandibular joint hypermobility in classical singing

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ABSTRACT: In classical singing pedagogy, jaw opening is extremely important for the blending of registers. However, controversial opinions amongst singing teachers still exist concerning the benefits of the traditionally advised “dropped rather far” jaw to maintain a low larynx position and freedom in singing. To those singers who have temporomandibular dysfunctions (e.g. hypermobility), the management of this hung jaw espoused in voice training might be difficult and lead to long time discomfort and pain. The purpose of this study is to investigate the effects of the use of two unitary orthodontic buttons placed on the superior and inferior back teeth with a string limiting its opening on general voice quality and laryngeal function during singing, used as a possible coping strategy for singers who present temporomandibular joint hypermobility. One classically trained singer diagnosed with temporomandibular joint hypermobility was recorded during the performance of different singing tasks by means of electrolaryngography, when using and not using these unitary orthodontic buttons. Preliminary results on the effects of the use of this strategy on acoustical and laryngeal parameters suggest that unitary orthodontic buttons might be used by classically trained singers who suffer from temporomandibular joint hypermobility as a copying strategy to avoid discomfort when opening wide the mouth to sing high notes. This seems to contribute to acquire optimal jaw opening control without compromising overall general voice quality. Moreover, the use of this aid seems to increase the average closed contact of the vocal folds during phonatory cycles, suggesting additional pedagogical benefits. Further research in this field should be developed, assessing the prevalence of temporomandibular joint hypermobility amongst classically trained singers and associated impacts on their singing technique and quality of life.

1 INTRODUCTION

In classical training pedagogy, singers are trained to be acoustically more efficient by means of increasing vocal loudness (to be heard with the orchestral accompaniment) but with minimum laryngeal effort associated (Howard, 1995). This can be achieved by a combination of several factors, including the manipulation of a low larynx position, regardless of vowel and pitch (Miller, 2000). It is known that changes in larynx height are associated with changes in the length of the vocal tract, and thus on voice timbre and acoustical projection (i.e. changes in formant frequency distributions) (Sundberg, 1987). A high larynx position is associated with shortening of the vocal tract, whereas a low larynx position is associated with its lengthening, an important phenomena used to improve the ability of the vocal tract to transfer sound (Sundberg, 1987). Therefore, singing pedagogues tend to teach their students different articulatory habits as those used for normal

speech, since for normal speech there is a natural raising of the larynx with increasing phonation frequency (Sundberg, 1987).

Singers learn to maintain a low jaw position in order to maintain a low larynx position (as larynx and mandible are connected by the hyoid bone and respective attached muscles), regardless the vowel and throughout their entire vocal range. Thus, a low jaw position tends to corresponds to a low larynx position, especially for high sung tunes (Cookman & Verdolini, 1999).

Classical singers are generally encouraged to drop their jaw when approaching “passagio” notes (i.e. specific frequencies for which there are significant changes in laryngeal mechanisms involved in voice production) (Miller, 2000), or when singing high notes, in order to counteract the natural tendency of raising the larynx and diminish the size of the pharynx (Rose, 1971). Singing teachers seem to agree that there is a correlation between jaw opening and laryngeal activity. However, controversy still exists towards different jaw opening approaches

during classical singing. On the one hand there are pedagogues, such those belonging to the English school of vocal pedagogy, that teach a drop rather far jaw position in order to achieve an "open throat and freedom in singing" (Rose, 1971). On the other hand, other pedagogues believe that the position of the mouth or jaw should be highly dependent on the vowel, consonant, tessitura and intensity level, and thus there is no ideal position for mandibular movement (Miller, 2000). And finally, there are those pedagogues that believe that a "relaxed jaw, lower larynx and relaxed vocal membranes" are beneficial approaches for training theatre voices (Lessac, 1967). This controversy might exist due to a lack of available data. The results of a recent study by Cookman and Verdolini (1999) suggested that, for conversational pitch, extreme mouth opening seem not to be beneficial as it limits laryngeal adduction; however, no conclusive results were obtained for the relationship between jaw opening and laryngeal activity during singing high pitches (Cookman & Verdolini, 1999).

There is an associated increased risk of temporomandibular joint hypermobility development with frequent singing (Sataloff, Brandfonbrener & Lederman, 1998), so it is possible that classical singers might be more prone to develop temporomandibular joint hypermobility. Therefore, it would be important to investigate whether singers suffering from temporomandibular joint hypermobility could benefit from the use of two unitary orthodontic buttons placed on the superior and inferior first molar teeth with a string limiting jaw opening as a copying strategy, and assess its implications on overall quality of voice.

Temporomandibular joint hypermobility is a common disorder within the general population, affecting mostly young adults and the female population (Sataloff, Brandfonbrener & Lederman, 1998). According to previous studies, approximately 5% of the general population is highly affected with pain related to temporomandibular joint hypermobility. Because of the greater incidence of this condition amongst the female gender, it is commonly associated with psychosocial stressors (Rugh & Solberg, 1985). Taking into account the potential stresses and demands associated with classical singing (number of practicing hours and rehearsals, acoustic demands of different venues, types of repertoire and of audience) (Sataloff, Brandfonbrener & Lederman, 1998), it is probably that the incidence of TMJ dysfunctions amongst singers is higher in comparison to the rest of the population. Epidemiological studies comparing the prevalence of temporomandibular joint hypermobility symptoms between flautists and the general population suggest similar trends between the two groups, with bruxism being more

prevalent in flutists (44% flutists presented teeth grinding compared with 19% of the general population) (Baken, 1992). However, few studies have been concerned with the prevalence of TMJ disorders amongst classical singers, especially when it has been suggested that an aggravating factor of joint laxity or hypermobility is frequent singing (Sataloff, Brandfonbrener & Lederman, 1998).

As presented above, there is an ongoing discussion amongst singing teachers pedagogues on the interactions of jaw opening and laryngeal mechanisms, thus on the effects of jaw opening on voice quality. During the current pilot study, one of the first observations when recruiting participants was the prevalence of temporomandibular joint hypermobility amongst all of the approached singers (6 singers in total, 5 women and 1 man). Therefore, the aim of this study is to understand more deeply these interactions between mandibular mobility and laryngeal mechanisms, as well as exploring the use of two unitary orthodontic buttons with a string limiting jaw opening, as a pedagogical tool to be used in the future with those singers who present pain associated with temporomandibular joint hypermobility.

Orthodontic buttons attached to the facial surface of the canine teeth connected with an interarch elastic band on each side of the mouth have been referred as an inexpensive aid for the management of temporomandibular joint hypermobility (Sataloff, Brandfonbrener & Lederman, 1998). Thus, this pilot study aims to explore the application of a similar strategy (i.e. two stainless-steel mesh-based buttons) for the optimization of the singing performance for those who suffer from temporomandibular joint hypermobility and associated symptoms.

The hypothesis to be tested are: (i) the use of two unitary orthodontic buttons to limit jaw opening can be used as an aiding tool to correct temporomandibular joint hypermobility without interfering with overall sound quality; (ii) the use of this orthodontic buttons might offer pedagogical benefits to re-educate jaw positions and singing technique for those singers suffering from temporomandibular joint hypermobility; (iii) based on previous results (Cookman & Verdolini, 1999), it is possible that laryngeal adduction is decreased with lower jaw positions during singing.

2 METHODS

2.1 Participant

A young healthy classically trained singing student (age 26 years old), classified as a light coloratura soprano, who has been confirmed to

have temporomandibular joint hypermobility (by means of lateral transcranial radiography, presented in Figures 1 and 2, showing on the left side the condyle far beyond the fossa; we are also able to compare the normal joint in the right side, where there is no displacement of the condyle). The participant was having regular singing lessons (i.e. one hour once a week), volunteered for the study. The voice was perceptually assessed by a singing teacher as being normal on the day of the experiment and no history of voice disorders until date was reported.

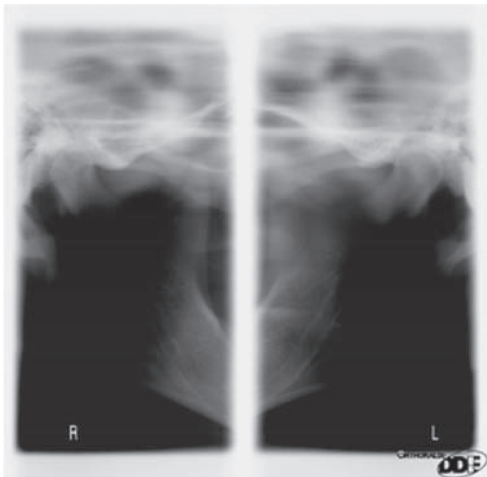


Figure 1. Transcranial radiography of the TMJ with a closed jaw position.

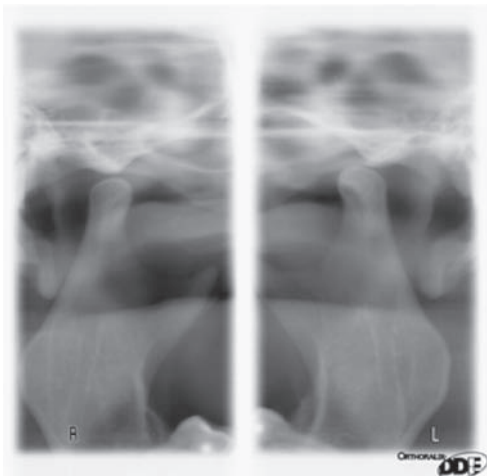


Figure 2. Transcranial radiography of the TMJ for maximum jaw opening.



Figure 3. Two unitary orthodontic buttons with a steal string limiting maximum jaw opening.

2.2 Design

This case study involved electrolaryngographic recordings during the performance of different singing tasks, under two different experimental set-ups: (i) non limited jaw opening (maximum mouth opening 52 mm); and (ii) limited jaw opening, by means of two unitary orthodontic buttons placed on the superior and inferior back teeth [first molars (16, 26, 36, 46, according to the SDI nomenclature)] with a string in between (maximum mouth opening 33 mm) (Figure 3).

2.3 Procedure

The electrolaryngographic recordings were carried out with the new digital Laryngograph® (Laryngograph Ltd.), which consists of a laryngograph processor, connected to two plated electrodes, an omnidirectional (pressure sensitive) wide band flat response EK3132 electret condenser series microphone, with ± 2 dB 100 Hz to 10 kHz noise level 26 dB (SPLA), and dynamic range of 88 dB, and linked via a USB interface to a portable Pentium-M (Centrino) processor, 512 MB RAM and 100GB IDE HDD. The electrodes were positioned at the level of the larynx and their position confirmed with reference to the output waveform (Lx) displayed on the screen of the laptop. The head-mounted omnidirectional (pressure sensitive) electret microphone was placed at the top left side of the front head, with a distance of 15,5 cm away from the left corner of the mouth. The calibration was done according to the procedure mentioned on the Laryngograph's Speech Studie Users Guide.

The electrolaryngograph (Figure 4) was chosen for this study because it allows aspects of vocal fold vibration to be monitored, such as vocal fold adduction. The electrolaryngograph generates a signal that does not have components of supra-glottal



Figure 4. Image of the electrolaryngograph used during voice recordings (left), showing a detail of the electrodes² and microphone position meanwhile performing the singing tasks (right).

influence, thus providing accurate information about vocal phenomena (Baken, 1992). Additionally, it is a non-invasive means of assessing aspects of vocal fold vibration, thus not interfering with singing and the performance of classical repertoire (Carlson & Miller, 1998).

For the purposes of this study, the participant was asked to: (i) sustain the vowel [a] on five different pitches, namely A_3 , E_4 , B_4 , F_5 and Bb_5 ; these particularly pitches were chosen since, according to previous research on training soprano voices (Miller, 2000), seem to reflect different vocal mechanisms and laryngeal and vocal tract adjustments; (ii) sing a glissando (i.e. when the whole singing vocal range is explored) using the whole range of the voice, in both forte and piano conditions; (iii) sing the first forty-six bars of the operatic aria “Der Hölle rache kocht in meinem Herzen” from the Opera “Die Zauberflöte”, by W.A. Mozart, involving a two octaves vocal extension of (i.e. F_4 to F_6);

These singing tasks were performed first without the use of two unitary orthodontic buttons with a string limiting mandibular opening (maximum jaw overture of 52 mm), and secondly with the use of this orthodontic buttons limiting jaw opening (maximum overture of 33 mm). The length of the string of the two unitary buttons was chosen according to what it was comfortable for the singer in terms of jaw opening, and ability to sing the different pitches, and taking into account the application of this orthodontic tool on other situations (Sataloff, Brandfonbrener & Lederman, 1998). These frequencies were first played in a keyboard synthesizer and then sung by the singer.

2.4 Vocal analysis

Data collected for the different singing tasks were analyzed using a laryngograph micro-processor, a laryngograph PC interface card (PCLX), a laryngograph PCLX software—‘Speech studio’ (Spead and Qanalysis) and a compatible portable

PC computer. The sampling rate used for this study was 22050 Hz, according to the vocal range that would be analysed. ‘Speech studio’ enabled the display, analysis and printout of the laryngographic waveform and corresponding fundamental frequencies (F_x), closed quotient ratio (Q_x) and amplitude of fundamental frequencies (A_x), as means of a range of different graphs. These allow quantitative analyses of important aspects of voice production [i.e. pitch (related to F_x analysis), vibratory pattern of vocal folds vibration (related to Q_x analysis), and loudness (related to A_x analysis)]. Additionally, the program offers a range of quantitative data on acoustical aspects of voice production for sustained vowels (e.g. harmonics-to-noise ratio) that can be statistically compared.

To test the first research hypothesis, acoustical analysis were performed using long-term average spectrum (LTAS), which was calculated for the two conditions of the study (i.e. with and without jaw limiting), for the excerpt of the operatic aria. LTAS analysis was particularly chosen because is a measure of general voice quality, providing that the sample is long enough (which was the case of the operatic aria) (Leino, 1993). By comparing LTAS for both conditions, it is possible to assess whether overall voice quality is affected by the use of orthodontic jaw opening limiting buttons.

To test whether the placement of the two unitary orthodontic buttons could affect laryngeal activity by limiting jaw opening, and thus affecting vocal tract properties, changes of sound pressure level (SPL), which reflect dynamic properties of both voice source and vocal tract properties in relation to fundamental frequency (F_0), were also assessed by means of a phonetogram (Gramming, 1988). As used by previous studies (Leino, 1993), the vocal task used was singing a glissando to compare the voice in its whole dynamic range for the two conditions of the study. To assess the relationship between jaw opening and laryngeal activity during singing, vocal analysis were focussed on parameters that could reflect vocal fold abduction and adduction, namely closed phase quotient (Q_x) (i.e. the time for which the vocal folds are together per cycle, expressed as a percentage of the cycle’s period) (Howard, Lindsey & Allen, 1995), an its relationship with F_0 , for the operatic aria excerpt.

3 RESULTS

Comparisons between LTAS for the two conditions of the study (i.e. non limited jaw opening and limited jaw opening) and the performance of the excerpt of the operatic aria suggest that the placement of the orthodontic buttons does not

interfere with overall voice quality. As it can be seen in Figure 5, both LTAS (with and without jaw limitation) are extremely similar, thus suggesting no interference with the overall acoustical singing quality for this singer presenting temporomandibular joint hypermobility.

The dynamic properties of both voice source and vocal tract properties in relation to F0 seem to be different for the two conditions of the study. Differences between phonetogram displays for the glissando singing task can be observed (Figure 6).

When the singer sung the glissando using the orthodontic buttons limiting jaw opening to a maximum of 33 mm, the relationship between Qx and Fx is more linear and presents less scattered points in comparison to the phonetogram display of the same singing task performed when jaw opening was not limited. Additionally, it can be seen that vocal extension seemed to be wider for the limiting jaw opening set-up, with better defined contours (highlighted with a red circle in the phonetogram) for the high pitch frequencies.

Concerning vocal fold adduction, comparisons of both scattergrams of Qx against log (Fx) for the sung excerpt of the operatic aria suggest that

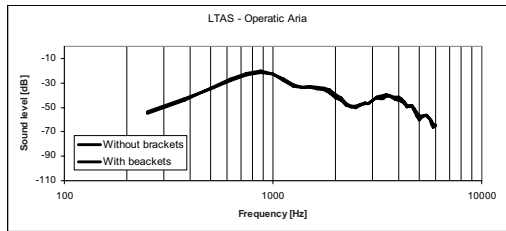


Figure 5. Comparison of LTAS for the 46 first bars of the operatic aria “Der Holle rache kocht in meinem Herzen” from the Opera “Die Zauberflöte”, by W.A. Mozart, with (red) and without (blue) orthodontic brackets limiting jaw opening. Maximum jaw opening without limitation is 52 mm for the highest note (F_0); maximum jaw opening with limitation is 33 mm for the highest note (F_0).

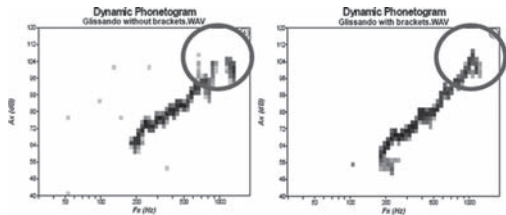


Figure 6. Plots of vocal intensity (Ax) against the logarithm of fundamental frequency (Fx), displayed as a dynamic phonetogram, for the sang glissando, and for the two conditions of the study: (i) with no jaw limitation (left) and (ii) with jaw limitation (right).

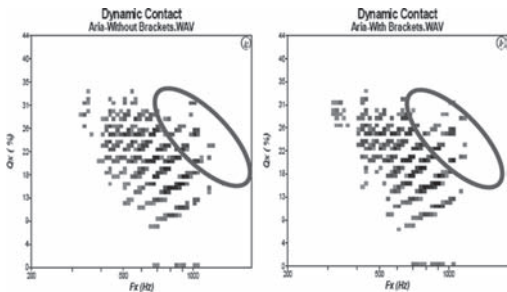


Figure 7. Dynamic contact scattergram of closed quotient expressed in percentages (Qx) against logarithm of fundamental frequency (Fx), showing that for the sung excerpt of the aria, closed phase contact seems to be longer when using the two unitary orthodontic brackets, especially concerning higher sung frequencies (marked with a red circle).

when jaw opening is limited, closed phase quotient becomes longer for higher pitch frequencies. This can be seen by the higher concentration of data points in the corresponding dynamic contact scattergram (highlighted with a red circle in Figure 7).

4 DISCUSSION

Although this investigation was exploratory and the results only preliminary, it seems that the use of two unitary orthodontic buttons as a copying strategy for singers who suffer from temporomandibular joint hypermobility might offer some healthy and pedagogical benefits, highlighting the necessity for further research to be developed.

LTAS comparisons for both limiting and non limiting jaw conditions suggest that the use of this strategy seems not to interfere with overall voice quality, as differences were not evident. Thus, it is possible that classical singers could use this orthodontic aid as a tool of temporomandibular joint hypermobility correction without compromising voice sound quality. On the contrary, the results suggest that vocal tract control and related laryngeal activity might benefit from the use of this orthodontic tool, as the phonetogram displays of the glissando task were more linear when limiting jaw opening. The wide F0 range demanded by classical singing implies the use of different vibratory modes, and it seems that the traditionally taught pedagogy of ‘drop back jaw’ might not always be beneficial to the laryngeal adjustments that happen throughout the whole vocal range. This might be especially the case for singers who suffer from temporomandibular joint hypermobility. Thus, it is possible that this copying strategy could be used as a pedagogical

tool to re-educate jaw opening habits and orofacial muscle activity for those singers who present significant temporomandibular joint hypermobility and associated discomfort when singing high frequencies.

According to previous studies (Howard, 1995), low larynx positions might be associated with relative abduction of the vocal fold (Zenker, 1964). Additionally a more acoustically efficient voice is obtained when the time during which the vocal folds are together per cycle is increased (Howard, Lindsey & Allen, 1995). These results were also observed in this study. A more linear variation of CQ with $\log(F_0)$ was obtained when using orthodontic buttons and higher pitches seem to correspond to higher Qx values for the jaw limiting set-up of the study (as seen in Figure 6). It is possible that smoother transitions in the laryngeal adjustments used to sing through the whole vocal range, and higher vocal efficiency are presented when jaw opening is limited towards an optimum position. Perhaps wider jaw openings might interfere with the optimum position of the larynx and larynx stabilisation when singing high notes and with its activity when laryngeal biomechanical adjustments occur through the vocal range.

However, at the moment these are only assumptions. It is important to highlight that, although these preliminary data suggest possible benefits of the use of unitary orthodontic buttons for singers with temporomandibular joint hypermobility, they should be carefully interpreted. Laryngeal mechanisms depend on a multitude of factors that were not taken into account during this study (e.g. subglottal pressure and laryngeal height). Additionally, changes in CQ with F_0 characterise just one of the many aspects of classically trained singing voice. Further research on the application of this aid into training classical voices should be undertaken, before solid conclusions can be drawn. It would be important to explore, in a much more controlled environment (e.g. using dual channel electroglottography to measure larynx height, using a higher number of singers, and following a comparative controlled study) acoustical and laryngeal properties. A further investigation is currently being undertaken, expecting that more robust and less exploratory data can be presented in future. With this study, we hope to contribute for a deeper understanding of the relationship between jaw position and laryngeal activity, and correlate that with perceptual analysis of voice quality undertaken by different vocal pedagogues, in order to discuss best approaches for mandibular overture for singers with temporomandibular joint

hypermobility. Additionally, another aim of this study is to assess the prevalence of this dysfunction amongst classically trained singers in comparison with other singing genres and the general population, to assess whether this could be a related condition to those who need to sing high vocal pitch ranges.

5 CONCLUSION

Frequent singing as been pointed out as an aggravating factor for temporomandibular joint mobility dysfunction (Sataloff, Brandfonbrener & Lederman, 1998). Additionally, singing high frequency ranges, such those existing in classical singing repertoire, usually involves wide jaw opening positions. This, for those singers suffering from severe temporomandibular joint hypermobility, might cause discomfort and impact negatively on the quality of their singing development. As facial pain highly interferes with conveying the beauty of the voice and emotional expression in artistic performance, dentists should have special attention with the singer's evaluation of temporomandibular joint hypermobility, and special pedagogical tools should be explored in future to contribute to the performance optimization of these singers.

Despite being only a preliminary case study, and presenting some methodological draw-backs, these data provided evidence that differences in vocal production can exist associated with the use of two unitary orthodontic buttons with a string limiting jaw opening. It is important that future research explores: (i) the prevalence of temporomandibular joint disorders amongst classical singers in comparison with other musicians and the general population; (ii) the distribution of temporomandibular joint dysfunctions amongst singers of different singing schools; and (iii) explore the use of orthodontic buttons with an interarch stainless-steel string as a pedagogical tool to re-educate jaw opening in singing for those singers suffering from temporomandibular joint hypermobility and associated symptoms.

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The biomechanical challenge with angled implants

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ABSTRACT: There must be a correct planning of the appropriate location and angulation of the dental implants. Nevertheless, anatomical conditions, bone morphology and aesthetics may determine another site for the implant bed and consequently a different angulation than the commonly accepted. In these cases, the forces applied to the implants may differ from the usual.

The longevity of implant-based prosthodontics depends mainly on how the masticatory forces are transferred to the implant and to the surrounding bone. The forces applied outside the center axis may induce overloading on the bone, however, the placement of angled implants avoids more invasive surgeries as the implants are placed in existing anatomical sites.

The authors present some clinical cases of oral rehabilitations with angled implants, approaching the biomechanical issue.

1 INTRODUCTION

1.1 Planning

A successful oral rehabilitation must be based on proper planning. In the case of treatments with implants, their number, location and angulation are of paramount importance.

The success of prosthetic restorations is directly related to the accuracy and passive adaptation between the prosthesis and the implant, obtaining good stability of the interface implant-abutment, as well as a proper distribution of the whole masticatory forces [1].

The precise adjustment between the implant and the supra-structure depends, from the start, on an accurate impression of the components' position in the oral cavity. Subsequently, it is essential to reproduce the clinical condition accurately in the work model. This is dependent on the type of material and technique of printing, and the material and technique of casting [2].

According to Pampel [3], a successful oral rehabilitation depends on the correct reproduction, in the laboratory, of the structures that form the basis for prosthetic support in the mouth, whether they are teeth or implants.

2 IMPRESSIONS

2.1 Materials

The material used to transfer the position of the implants in the mouth to the work model must

be sufficiently resilient—to be removed from the recess without distortion—and hard enough—to allow the settlement of the parts needed for the transfer and prevent the movement of components during casting.

As for the ideal characteristics, the material must have good strength, stiffness, elastic memory, resistance to permanent deformation and rupture and high degree of accuracy.

The materials that meet most of these features are elastomers which provide highly accurate impressions.

2.2 Techniques

The impression techniques can be divided into direct and indirect technique.

The indirect technique, also called reposition technique, involves the use of a closed tray and the transfer is retained to the implant at the end of the impression. Thus, the reposition of the transfer is crucial and the impression material is solely responsible for maintaining its correct position.

The direct technique, also called pick-up or drag technique, uses an open tray and the transfer is drawn with the impression. There is the need to remove the screw that holds the transfer, however, there are no intermediate steps that can cause inaccuracies.

Despite the indirect technique is clinically preferred, there are studies that indicate that transfers are not always placed correctly in the impression [4].

Carr [5] compared the direct and indirect techniques and demonstrated that the direct one is more precise.

3 ANGLED IMPLANTS

3.1 Clinical situations

Sometimes, the anatomical conditions, the bone morphology and aesthetics may determine the direction and angle of implants placement [6].

In the presence of insufficient bone volume, it is either necessary to perform advanced surgical procedures (more invasive), or the implants must be placed in angled positions. (Figures 1 and 2).

The unfavourable angulation of dental implants is a complication that can hinder the design and manufacture of the prosthesis [7].

The correction of the unfavourable position of the implants can be achieved through the use of angled pre-made abutments (until 20°) or custom ones [8].

However, in excessively angled implants, the prefabricated abutments do not solve the problem. In these cases, the solution is to make an impression to the implant level, even though it is a very difficult procedure.

3.2 Impressions in angled implants

When implants are placed in unfavourable positions or adverse angulations, a correct impression may be difficult and time consuming [9].



Figure 1. Orthopantomography showing an angled implant (upper left quadrant).



Figure 2. Reproduction of the situation in the oral cavity. Notice the angle of the distal implant.



Figure 3. Modified transfer to provide a right angle for the impression.



Figure 4. Cast with the modified transfer. Notice the difference of the angulation in comparison with figure 2.

The problem of angulation of implants and lack of parallelism hinders the impression technique due to the axis of tray removal, the possible rupture of the material and the extraction inertia.

Altering the transfer can be a solution to overcome the difficulties involved in the impression. (Figures 3 and 4).

4 FORCES

The longevity of implant-supported oral rehabilitations depends largely on how the masticatory forces are transferred to the implants and to the surrounding bone.

The factors that affect the load on dental implants are:

- Magnitude, direction and location of occlusal forces applied to the prosthesis;
- Type, geometry and rigidity of the prosthesis;
- Nature of the connections between the prosthesis and the implants;
- Number, location, angle, geometry, length and diameter of implants;

- Mechanical properties of the implants and the prosthetic restorations;
- Quality of the surrounding bone.

The off-axis forces may cause excessive load on the surrounding bone [10].

In a study of Canay [11], which compared the distribution of stress around angled implants with vertical implants, using the finite element analysis, it was possible to conclude that:

- On the horizontal loads there were not significant differences in the stress levels;
- For vertical loads, the values of compressive stress were five times higher.

5 CONCLUSION

The longevity of implant-based prosthodontics depends mainly on how the masticatory forces are transferred to the implant and to the surrounding bone.

The forces applied outside the center axis may induce overloading on the bone, however, the placement of angled implants avoids more invasive surgeries as the implants are placed in existing anatomical sites.

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Biomechanical and clinical performance of a cantilevered tooth-implant fixed bridge

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ABSTRACT: Tooth-implant supported fixed prosthesis present a potential biomechanical problem: the implant is rigidly connected to the alveolar bone and, by opposition, the tooth possesses periodontal ligament and greater amplitude of intrusion, which may lead to bone loss around the latter. The use of cantilever extensions is a common procedure in fixed rehabilitation, keeping its role in the therapeutic options of much of the clinicians, despite the fact that potentially destructive forces are transmitted to the abutments. These two options in oral rehabilitation are not consensual in scientific literature. This clinical case—tooth-implant supported bridge with mesial cantilever—reflects the expecting success once appropriate planning is established, using careful protocol and one is aware of the limits of each option to be taken. In this kind of prosthetic rehabilitation, to improve the performance and longevity of it, evidence-based biomechanical considerations are of the most important significance.

1 INTRODUCTION

The oral rehabilitation with fixed prosthodontics, besides being extremely wished by patients, is also a very efficient clinical resource at a functional and aesthetic point of view. Nevertheless, it is, like any therapeutic decision, not completely free from complications. The success depends on the case diagnosis, on the treatment options, on planning, on the technical execution and on preservation, having in mind, at the same time, that in any prosthetic decision, it is critical to consider the supporting tissues and occlusion. In fact, the clinical circumstance is always different and particular, requiring a specific treatment, sometimes distant from the abstract ideal, and demanding from the dentist not only a strong scientific and technical knowledge of the existing options, but also a skilled ability to establish adequate treatment plans—the art of choose the best for each one of our patients.

In fixed prosthodontics is quite common the need to use simultaneously tooth and implants as abutments and cantilevers. Treatment plans containing these resources present additional complications.

1.1 *Implant supported cantilever extensions*

Most of the complications that take place are related to the significant increase of the occlusal forces which are transmitted to the prosthesis, to the implant and to the marginal bone (Kunavisait et al. 2002, Romeo et al. 2003, Wennstrom et al. 2004, Bragger et al. 2001, Kim et al. 2005).

Concerning the prosthesis, these harmful forces result, after five years, in loss of cement (8,4%), ceramic fractures (3,5%) and infra-structure fractures (3%), being also frequently described the loss of adaptation or looseness in the thread of the screw which connects the prosthesis to the implant (Wennstrom et al. 2004, Pjetursson et al. 2007). Although one verifies there is no difference in the average amount of marginal bone loss in prosthesis over implants with and without cantilever, Wennstrom stresses, in 2004, that, in absolute values, there is a bone loss superior to 1 mm, in a five-years' period, in 33% of the cases, against 19% of the prosthesis over implants without cantilever. For Becker & Kaiser (2000) it is clinically wise to use, at least, two implants as abutments of fixed prosthesis with mesial cantilever.

1.2 *Combining tooth and implant*

Bragger et al. (2005) stated that the tooth-implant supported fixed bridges show more technical failures than the single crowns or the implant-supported prostheses. Nevertheless, Lindh (2008) points out there are no reasons for supporting the general belief that fixed prosthesis, combining teeth and implants, represent an inferior alternative to the rehabilitating treatments exclusively supported by implants. The complications depend on the direction and on the quantity of the occlusion strengths transmitted to the prosthesis (Wennstrom et al. 2004, Bragger et al. 2001, Kim et al. 2005, Menicucci et al. 2002). According to Pjetursson et al. (2007), among the technical



Figure 1. Initial radiographic view – 2002.

complications that occur in a five years' period, we find a fracture of the implant (0,8%), a fracture in the screw or in the abutment (0,6%), loose screw/abutment (6,9%), loss of retention (7,3%), adhesive/cohesive fracture of the ceramics (7,2%) and infra-structure fractures (1,6%). In the biological complications, the ones related to the soft tissues (7%) and the tooth intrusion (5,2%) stand out. This last one has been explained by an atrophy caused by the lack of use, the differential dissipation of energy, mandibular flexion and torsion, flexion of the infra-structures, by the ratchet effect, impaired rebound memory and debris impaction (Pjetursson et al. 2007, Ozcelik & Ersoy 2007, Pesun 1997).

In the case we wish to portray, we explore the context in which the treatment plan is put to action, discussing the orientations and cautions that have to be taken into attention when fulfilling a posterior oral rehabilitation using a tooth-implant supported bridge with mesial cantilever.

2 CLINICAL CASE

Female patient, 40 years old, healthy, with indication of fixed rehabilitation of the 1st quadrant (Kennedy's Class III), balancing the absence of both premolars.

Treatment Plan: Combined tooth-implant-supported fixed partial denture (FPD) with abutments in 16 (tooth), 15 (implant) and extension 14.

3 DISCUSSION

The following therapeutic options were taken into consideration:

1. Tooth-supported FPD on 13 and 16
2. Implant-supported FPD
 - 2.1 14 (implant) and 15 (implant)
 - 2.2 14 (implant) with distal cantilever extension
 - 2.3 15 (implant) with mesial cantilever extension

3. Tooth-implant-supported FPD
 - 3.1 14 (implant) and 16 (tooth)
 - 3.2 15 (implant) and 13 (tooth)
 - 3.3 16 (tooth) and 15 (implant) with mesial cantilever extension
 - 3.4 13 (tooth) and 14 (implant) with distal cantilever extension

When planning a fixed prosthetic rehabilitation, the first choice must be of a conventional prosthesis—an implant supported fixed-prosthesis or crowns over tooth or implant, leaving for a second choice (due to financial, anatomical or any other reasons centered in the patient preferences) the options of cantilever bridges supported by teeth, tooth-implant supported bridges or adhesive resin bridges (Pjetursson et al. 2007).

The assessment of the available bone quantity for placing the implants revealed an insufficient amount in the 14 region, having been decided, by the evident patient's will, not to make bone regeneration (the ideal option in terms of the tooth preservation) which made impossible the placing of an implant in this region. This fact invalidates options 2.1, 2.2, 3.1 and 3.4.

The causing factor common to the complications connected to these rehabilitations is the occlusal force resulting from the function, especially when it is of a non-axial component (Wennstrom et al. 2004, Bragger et al. 2001, Kim et al. 2005).

Considering this premise, the indications for the building up of occlusal schemes in rehabilitation with implants in the posterior zones, advise balanced contacts in occlusion of centric relation and maximum intercuspation, while in the disocclusion movements, it is advised to have contacts on the working side as much anterior as possible— anterior guidance or canine guidance (Kim et al. 2005). As a means of preserving the patient's canine-guided occlusion we have chosen not to involve this tooth in the rehabilitation to be done. This fact invalidates options 1 and 3.2.

It was left to decide about the use of the tooth 16 in a bridge with implant on 15 and cantilever on 14. We must take into account that, if there isn't enough scientific evidence to establish a protocol considering when teeth must be sacrificed and replaced by implants, there also isn't one for not prescribing fixed prosthodontics using, at the same time, teeth and implants (Lindh 2008). Besides this fact, Menicucci et al. (2002), defend that a transitory load applied for a short period of time to tooth-implant supported structures, doesn't cause a great stress. On the contrary, the tooth seems to work as a rigid structure, dividing its load with the implant, thus turning the bone stress into a much more homogeneous one. In this way, the periodontal ligament plays a key role in the long-time good

performance of tooth-implant fixed rehabilitations, considering that the viscoelastic resistance of the ligament is not overcome by exaggerated periods of time.

Thus, considering the options 2.3 and 3.3, we have chosen 3.3, that is, to include 16 in the bridge, for the following reasons:

- Raising the amount of abutments that support the prosthesis, improving the relation between the number of supports and the length of the bridge (Shillingburg et al. 1997).
- Keeping the physiological mastication and proprioception, the occlusion defense mechanisms and the control of the chewing forces by including a natural tooth with its periodontal ligament in the bridge and by the existence of natural teeth in the arch and also as oponents (Kim et al. 2005, Akça et al. 2006). A more favourable biomechanics, with a smaller lever movement—one of the most critical biomechanical factors for the survival of fixed rehabilitations, reducing the level of eventual loss of marginal bone related to the cantilever (Kim et al. 2005, Akça et al. 2006).
- The cantilever is mesial (Kim et al. 2005).
- The mesial abutment is an implant which causes less tooth intrusion (Becker & Kaiser 2000, Akça et al. 2006, Palmer et al. 2005).
- A bigger retention area in the distal abutment, reducing the unsealing risk (Palmer et al. 2005).
- The posterior abutment is a molar which means that it possesses the least mobility of all the teeth groups, due to the kind and intensity of the load, transmission of the strengths and biological reactions which, by themselves, are dependent on the size, area and shape of the roots (Pesun 1997, Shillingburg et al. 1997).

Considering all the initial conditionings, it is the authors' opinion that this combination of factors minimizes the potential risk of the chosen rehabilitation, advising it for this particular case, once the following technical impositions are respected:

- to decide for a cemented fixed prosthesis, thus allowing an improved passive adjustment and a considerable reduction of the stress concentration around the implants; (Taylor & Agar 2002)
- to establish a rigid connection between tooth and implant, reducing the stress concentration on this critical point of any rehabilitation (Taylor & Agar 2002, Palmer et al. 2005)

The re-evaluation, in 2007, five years after the setting of the fixed bridge, hasn't revealed any considerable bone loss, or any other of the complications that are usually described in the literature.

One has to point out that five years are enough for many of the probable problems to appear,



Figure 2. Final view—2002.



Figure 3. Re-evaluation in 2007.

particularly the ones that are more serious and compromising for the stability and longevity of the rehabilitation. Palmer et al. (2005), when referring to this topic, stress that many of the serious problems (fractures, loss of teeth and loss of vitality of the abutments) appear at 2 years period. We must also consider that the survival rate of tooth-implant supported fixed prosthesis reduces from 95,5% at five years to 77,8% at ten years, and the one of fixed prosthesis with cantilevers reduces from 91,5% at five years to 80,3% at 10 years (Pjetursson et al. 2007).

These results make it valuable to suppose that this combination might show signs of breaking down much earlier. Nevertheless, it is the authors' conviction that it is possible to control the predictable problems, foresee the critical areas and to provide solutions capable of keeping the stability and foresight of this kind of prosthetic work. Although it is speculative, in this particular case, and considering the execution of the mesial cantilever, the use of a tooth as a distal abutment (usually at



Figure 4. Radiographic re-evaluation in 2007.

the basis of problems related to the existence of a distal lever of strength) it seems to balance the structure and to favour the implant, sharing with it the stress and hindering the stress concentration in the implant's region close to the extension which is also improved by the short and rigid connection between the tooth and the implant, besides the other functional, biological and technical conditionings already described which improve the prognosis of this kind of rehabilitation.

4 CONCLUSION

Although it was controvert in the existing literature, the rehabilitation of this clinical case with a tooth-implant supported fixed partial prosthesis with mesial cantilever, performed in 2002, allowed us to keep the canine and the pre-existing canine-guidance and, on the other side, to avoid a bone regeneration surgery. This decision was based on the anatomical context and also on the patient's preferences, with a strict respect for the occlusion rules.

When it was assessed, in 2007, it showed clinical success, free of any complication, being rightful to expect that this may constitute a long lasting solution. Having in mind the available scientific literature, the tooth-implant supported rehabilitations must be considered with precaution, but must not be excluded from the treating plans in oral rehabilitation.

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Evaluation of the displacements transmitted to a pig jaw, by orthodontic and orthopaedic devices

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ABSTRACT: In this study it was evaluated the displacements patterns, induced in a pig jaw, by two orthopaedic and orthodontic devices, namely Disjuntor and Quad-helix. These devices are used in orthodontics, for the correction of maxillary athresia with posterior crossbite, through the opening of the mid-palatal suture (disjunction). The devices had been manufactured over a pig jaw and the field displacements were measured using ESPI technique.

1 INTRODUCTION

The maxillary expansion is an ancient method used in orthodontics, for the correction of the maxillary athresia with posterior crossbite, through the opening of the midpalatal suture (disjunction), using orthodontic-orthopaedic devices [1–8]. The expansion is achieved when a force is applied to the dental-alveolar structures, through a removable or fixed device, by exceeding the necessary limits for the expansion, anticipating the cellular reaction of the periodontal ligament and favoring the dissipation of the forces through the maxilla sutures [9–16]. Due to anatomic characteristics of the jaw and its resistance of several cranial-facial sutures (pterigo-palatal, to fronto-maxillary, to naso-maxillary and to zygomatic-maxillary), the expansion occurs in the horizontal plane with a pyramidal form, where the base is directed to the anterior inter-incisive region (anterior nasal spine), favoring the appearance of a maxillary midline diastema, and the vertex located posterior, in direction to the posterior nasal socket (posterior nasal spine), and consequently an inclination to vestibular of the alveolar processes and posterior teeth. In the frontal plane the disjunction of the jaw obeys to the same geometric configuration, with the

vertex located close to the fronto-nasal suture and with the base at the occlusal plane [8–12].

The clinical effects of Quadelix and Disjuntor devices have been widely studied [17–22]. However still exists some controversy about the effect of these orthopaedics devices, related to the amount of force applied by each one and the displacements field induced by them in the surrounding maxillary structures and skull sutures.

The objective of the present study was to compare the displacement fields induced by two different orthodontic and orthopedic devices, Quad-helix and Disjuntor, in a dry pig jaw.

2 EXPERIMENTAL PROCEDURE

The two devices have been, firstly, manufactured over the model of a pig jaw. The Quad-helix device was manufactured with wire 0.9 mm/36 Dart/Wire from Remaloy® of Dentaaurum, and the Disjuntor device was made with a Hyrax screw 12 mm of the Forestedente® (Figure 1).

The applied load for the disjuntor was made twirling the screw Hirax. For quad-helix device the load was applied heating the wire. In this manner, it was possible to make a more uniform load,



Figure 1. View of the two orthodontic and orthopaedic devices, applied in a pig jaw.

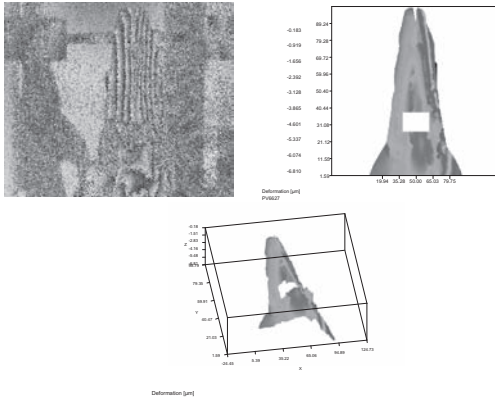


Figure 2. Displacement filed induced by Disjunctur device, for an applied load corresponding to a 1/4 turn of screw Hirax.

because in a recent study it was verified that the applied load depends a lot on the degree of manual activation, made by the clinicians.

The measurements were made using ESPI technique (Electronic Speckle Pattern Interferometry) [31, 32]. This technique allows the measurement of the displacement field in the perpendicular direction to the surface of models. setup used within this study.

3 RESULTS AND DISCUSSION

In Figure 2 are displayed the results obtained for disjunctur, with an applied load corresponding to a 1/4 turn of screw Hirax. It was observed for this device, that the displacement is higher in the region of posterior teeth, where the device is anchored at teeth by bands, and also in the anterior region, corresponding to the pre-maxillae.

Looking to phase maps, it was observed a symmetrical displacements distribution. The same pattern was also observed for higher applied loads. Take this into account, it is possible to say that for this device there is repeatability of results.

For quad-helix device it wasn't observed a symmetrical distribution of displacements in the jaw, as

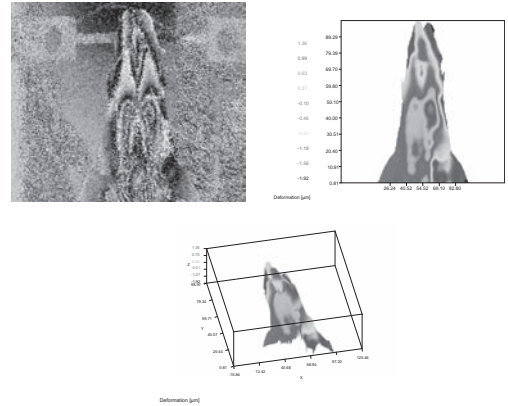


Figure 3. Displacement filed induced by Quad-helix, for an applied load corresponding to the heating of the wire.

it can be seen in phase map of Figure 3. It can also be verified that the displacements are more uniform.

For the two devices it has also been determined that, besides the displacement in the horizontal plane, especially on the posterior teeth, there is also a labial movement of the posterior teeth and alveolar processes, and a back rotation of the anterior section of the jaw, as already was verified by some authors [17–22].

It was also evaluated the displacement fields in the skull and it was verified that the two devices induces displacements in that region. It was also observed the same differences in the behavior of displacements.

4 CONCLUSION

With this work he was demonstrated that with the device disjunctur the applied load is greater than with quad-helix and as a result, the observed displacements are greater for the first one. It was also observed that the use of orthodontic or orthopaedic devices induces displacements in entire skull. These two devices provoke a rotation for vestibular position of posterior teeth and alveolar processes.

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Comparison of bracket/arch friction for passive self-ligated and conventional brackets

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ABSTRACT: The current challenges concerning systems of brackets available are, among others, achieving tooth position control during mechanical sliding, associated with low coefficients of friction.

With this study it is intend to compare the friction generated between two types of brackets: passive self-ligating (Damon 3MX,Ormco Corp., Glendora, Calif.) and conventional (Orthos,Ormco Corp., Glendora, Calif.) with elastic ligatures (Ormoplast Gris,Ormco Corp., Glendora, Calif.). For this purpose, it will be used two experimental models of a maxillary dental arch in pink acrylic resin, with six brackets (six anterior teeth) cemented with cianoacrylate (Super 2000). In order to measure the force that overcome friction, between arc and bracket, a dynamometer was applied.

1 INTRODUCTION

In 2005, Keim (Editor of the Journal of Clinical Orthodontics), referring to the American Association of Orthodontists Annual Session, said [1]:

This year, (...) there were three major topics bonding change for the future: three-dimensional imaging, endosseous skeletal anchorage, and self-ligation. (...) I've always been a little skeptical of claims of life-altering experiences, but the self-ligating systems I have actually tried seem work quite well.

However, Keim says that scientific evidence is needed before we clearly accept the manufacturers claim on the advantages of self-ligating brackets systems (SLB).

Herradine [2], in a study on the effectiveness of treatment with passive self-ligating brackets (PSLB) Damon SL (Ormco, Glendora, Calif.) concludes that there is an average reduction of 4 months in active treatment time and are held, on average, 4 appointments less when compared with conventional brackets (CB) [2]. This study, corroborates with other's, supporting the use of SLB with significant clinical benefits, this has raised the challenge to seek lower treatment times [3].

Miles et al [4], developed an in-vivo study with 58 patients that compared the use of Damon 2 SL (Ormco, Glendora, Calif.) with CB and concluded that patients preferred the aesthetics of CB, got

higher failure rate with PSLB, and, although the lower initial pain (arc round 14 thousand Copper NiTi,Ormco) reported by the patients, it increased substantially with the placement of a rectangular arch 16 × 25 mil (Copper NiTi,Ormco). According to this author, the differences between this and previous studies may be due to: decrease in treatment time may be influenced by more efficient techniques instead of only the use of PSLB, patients with severe crowding and indication for extraction may react more positively to PSLB and, the decrease in treatment time may be due to different mechanics or unintentional errors [4].

The friction force (FR) is the force that resists the movement of a body over another and is in the opposite direction to the movement. This force is proportional to normal force (FN) that acts perpendicular to the opposite direction of the movement. The friction resistance (RS) is characterized by the Static friction force (FRs), or required force to produce the initial motion and Dynamic friction force (FRk), which maintains the body in motion.

Factors that influence friction in orthodontics include: material and design of the bracket, material and shape of the arch, ligation method, and factors related to the patient, such as brackets/arch angulation and torque, dynamic forces and organic films [5–8].

The Damon Philosophy lies on the principle that only the minimum force necessary to start the

tooth movement should be used in order to prevent vessels occlusion of periodontal membrane, allowing the cells and chemical messengers responsible for dental movement to be transported to the local area [9].

The ideal orthodontic treatment combines forces capable of increasing the rate of tooth movement with minimum damage to the root, periodontal ligament (PDL) and alveolar bone [10].

Ong and Wang [11], following studies by Ericsson et al. [12], indicated that high orthodontic forces outweigh vessels blood pressure, leading to compression of the PDL in the pressure zone with consequent hyalinization and reduction of tooth movement, on other hand, light and continuous forces can cause mild ischemia with bone resorption/formation and simultaneous increase in tooth movement. The continuous tooth movement, without occlusion of blood capillaries, reduces the risk of bone loss in subjects with reduced bone support [11, 12].

The magnitude of force applied with conventional straight wire techniques is usually high (up to 500 grams Force), which can lead to tissue necrosis and degeneration after and between each activation, as it exceeds the response capacity of LPD [10, 13].

Currently, orthodontists are faced with two conflicting requirements regarding the control of tooth position. It is accepted that for a tooth to slid along an arch with minimum resistance, during the initial alignment, space opening or closing, one should de in the presence of low friction, i.e., lower possible contact arch/bracket; a close arch/brackets contact is required in order to obtain precise tooth movement control (torque, angle and rotation), which is associated with high friction, not suited to the movements of dental sliding.

The root axial inclinations control during orthodontic treatment providing appropriate root parallelism with regular interdental bone distribution allows obtaining and maintaining a stable result. In clinical practice, this control results from the third-order angulation (torque) that is printed in rectangular section arch of appropriate section during space closure and anterior root correction. Furthermore, arches of larger section, which fill the slot, are used in order to express third order brackets prescription. This process prolongs treatment time and increases the concentration of forces in the apical portion of the root [14]. Despite the excellent in-vitro SLB results with round arches of lesser caliber, used in the initial phase of treatment (alignment and leveling), when arches of greater size with rectangular section are used not greater differences were found between SLB and CB [5].

2 EXPERIMENTAL PROCEDURE

2.1 Materials and Methods

For this study, two experimental models of a maxillary dental arch in pink acrylic resin were made, as base, Figure 1, in which were cemented six anterior brackets with cyanoacrilate (Super 2000) (1.3 to 2.3 teeth), Figure 2, with the guidance of an 21 × 28 mil stainless steel arch obtaining a passive bracket bond.

Both models have all teeth ideally aligned, except the right lateral incisor (1.2) that is palatine to the remaining teeth.

In model 1 SLB (Damon 3 MX,Ormco, Glendora, Calif.) and in model 2 CB (Orthos,Ormco, Glendora, Calif.) with elastic ligatures (Sani-Ties Blue, Int GAC, Bohemia, NY) were

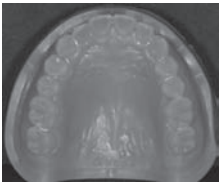


Figure 1. Acrylic resin experimental model.

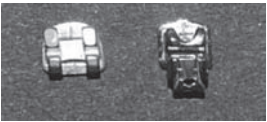


Figure 2. CB tooth 11(Orthos, Ormco, Glendora, Calif.) and PALB tooth 11 (Damon 3MX, Ormco, Glendora, Calif.).

Table 1. List of materials.

Product	Design	Dimension	Material	Manufacture
Brackets				
Damon 3 MX	Slide	22 mil*	SS†	Ormco, Glendora, Calif.
Orthos	Conventional	22 mil*	SS†	Ormco, Glendora, Calif.
Arches				
True arch	Rectangular	19 × 25 mil*	SS†	GAC Int., Bohemia, NY
Elastic ligatures				
Sani-Ties Ring Blue			Elasto-meric	GAC Int., Bohemia, NY

* 1 mil = 0.001 in = 0.0254 mm.

†SS = stainless steel.

used in the tests. The orthodontic arch wire used in both models was 17 × 25 mil stainless steel pre-fabricated (Preformed Arch-Wire True Arch, GAC International, Bohemia, NY).

The cleaning of each set arch/bracket was made with ethanol (95%) and air dried in order to remove impurities at the interface arch/bracket.

To get the value of maximum load or force necessary to overcome friction (FRs), each set arch/bracket is mounted on a dynamometer (KERN CH15K20) with a 0,005 kg/Force reading error, and a traction load is applied with a 200 milliseconds (ms) frequency of measurement in order to move the arch along the bracket for a total of 8 millimeters (mm). The scale was calibrated at the beginning of the experimental work and between each test.

Each test is performed in the dry state at room temperature (22°C).

Each test is performed five times, to obtain a mean value. This allows debugging of sporadic reading and verifying the repeatability of the system.

Five tests were carried out (test 1 to 5, Table 2) for each combination:

- 1. BC/ESQ: conventional brackets with traction performed on the left side of the model.
- 2. BALP/ESQ: self-ligating brackets with traction performed on the left side of the model.

A total of 25 tests were made.

Elastics were placed at the beginning of each new test.

3 RESULTS AND DISCUSSION

3.1 Results

In the presence of an ideal arch, for the Damon 3 MX brackets the mean force needed to over-

come the friction is virtually zero (T1 = 0.04 KgF), unlike the Orthos, brackets whose mean force (Fm) reaches 0.48 KgF.

There are no significant differences in the Fm value for PSLB as it increases the number of aligned brackets included in the arch, contrary to what happens to CB where there is an increase of 91% of the force necessary to overcome friction (T5 KgF = 0.04 vs T1 = 0.48 KgF). We can also observe that the same force is needed in the case of 1 CB (T5 = 0.04 KgF) and for 5 PSLB (T1 = 0.04 KgF).

The inclusion of a misaligned tooth (1.2) on the arch causes a significant increase in the average strength of around 97% (T1 = 0.04 KgF vs. T2 = 1.38 KgF) for the PSLB. This difference is significantly greater than in the case of CB where there is an increase of only 27% (T1 = 0.48 KgF vs. T2 = 0.66 KgF).

When comparing the two systems with six brackets included in the arch, we see that the Fm needed to overcome the friction for the PSLB (T2 = 1.38 KgF) is two times higher CB (T2 = 0.66 KgF) in the presence of a misaligned tooth.

A positive relationship was found between increasing the number of brackets aligned, in the presence of a misaligned bracket, and Fm for CB (E3 = 0.18 KgF vs E2 = 0.66 KgF), which does not occur for PSLB (E3 = E2 = 1.38 KgF).

3.2 Discussion

The results found are in agreement with previous studies with pre-formed arches, which refer a significant increase in the value of Fm necessary to overcome the friction in the presence of one or more misaligned brackets [15,16]. A study by Henao and Kusy, with pre-formed arches 19 × 25 mil SS with maxillary typodonts were seven brackets and two tubes molars where cemented obtained Fm values between 0.47 and 2.70 KgF for PSLB (Damon 2 SL, Ormco) and 0.64 to 3.25 KgF for CB [16]. These values are in agreement with those obtained in this study.

The values of Fm in the passive and active state for PSLB, observed in this study are in agreement with several previous studies that found Fm in passive state ranges between 0.032 and 0.065 KgF, depending on the degree of malocclusion, so the system forces in three dimensions, not only the ligation force, influences the FR at the arch/bracket interface [17–21].

We observed significantly lower Fm values for CB both in the passive and active as in the active state for CB comparing to values previously reported [18,22]. This may be due to the type of elastomeric ligature and section of the arch used,

Table 2. Laboratorial tests.

Tests (T)	Brackets
Test 1	13, 11, 21, 22, 23
Test 2	13, 12, 11, 21, 22, 23
Test 3	12, 13
Test 4	12
Test 5	22

Table 3. Mean values of force (Fm) expressed in KgF.

Test	BC/left	BALP/left
T1	0,48	0,04
T2	0,66	0,38
T3	0,18	0,38
T4	0,12	0,60
T5	0,04	0,02

as in one of the studies that was 19×25 mil SS arch. Moreover, the proliferation of SLB and the increasing of knowledge on the biology of orthodontic movement may also have led to the improvement of CB and elastic ligatures by orthodontic materials manufactures, in order to contribute to the decrease of FR.

As in a study by Tecco, we observed that Fm rises more significantly in PSLB with rectangular section arches, then in CB, only in the presence of malocclusion. This, emphasizes the importance of alignment before using arches of greater size in the sliding mechanics. Moreover, contrary to what Tecco concludes in his study, we found that there are significant differences in the Fm values depending on the number of CB and PSLB in active state involved in the test, apparently, the same relationship is not true for PSLB in passive state [23].

4 CONCLUSION

The average force needed to overcome friction in the presence of a misaligned tooth increases, substantially more, in the case of passive self-ligating brackets compared to conventional brackets and is too times higher for the upper anterior sextant Antero with a misaligned upper lateral incisor using a PSLB system.

Forces applied to the teeth are transmitted by the deflection of the arch, so in the passive state (ideally aligned arch) deflection does not occur, and so, there aren't any forces applied to the tooth. In the presence of an active state (such as with a misaligned tooth) the deflection of the arc creates normal force and classic FR, both in CB and PSLB, thus malocclusion and not only the type of bracket, has a significant impact on FR.

Friction influence all phases of orthodontic treatment, the clinician must review the treatment goals before selecting a bracket system. Although it is desirable to obtain minimum FR during initial treatment (alignment phase), this may contribute to a decrease in the control of root position at later stages.

Moreover, before the beginning of sliding mechanics with rectangular section arches, that allow greater control of the root position, the arch must be previously aligned and leveled to avoid the exponential increase of the mean force required to overcome friction, both for PSLB as for CB. In case of PSLB, that care must be doubled, since the effect on increasing the Fm value is substantially greater.

In the future will be required more in-vivo studies that compare the effectiveness and efficiency of the PSLB and CB.

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Evaluation of the quantity of MMA released by denture base resins

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ABSTRACT: The aims of this study were to evaluate if the temperature, time elapsed after preparation and different types of solvent influence the release of MMA from a self-curing resin. The measurements were performed by spectrophotometry. The monomer was release to the solvent in all experimental settings. No statistically significant differences were found for tested temperatures or for different solvents. It was observed that waiting 24 hours after preparing the resin reduces considerably the amount of MMA released.

1 INTRODUCTION

The use of PMMA self-curing resins, were for the manufacturing of removable partial dental prostheses, orthodontic devices, artificial teeth and other kind of dental restorations [1, 2]. To make in laboratory the acrylic resin is necessary to use a powder, corresponding to the PMMA (polimethyl methacrylate), modified by small amounts of ethyl, buthyl and other methacrylates are necessary to increase the impact resistance of polymer. The liquid contains MMA (methyl methacrylate), with inhibitors in small amounts, to enlarge time life of monomer [1].

During polymerization, the chemical reaction is incomplete, leading to the release of MMA constantly [3–6]. Self-curing acrylic resins release more MMA than those thermopolymerized [7–10].

Some factors that influence the release of MMA are as such powder/liquid Ratio; polymerization time; type of polymerization; pressure [6, 8, 10–13].

The released of residual monomer during the incomplete polymerization reaction, is known to have a cytotoxic effect, that can cause injuries to the oral mucosa [2, 6]. It is described, some kinds of inflammatory reactions of the soft tissues as sensitivity, allergic reactions, eczema, edema. [4, 7, 14].

It is also reported respiratory difficulties, heat in mouth, after the prosthesis being settled in mouth [15]. There are histological evidences of lymphocitary activity, destruction of cellular membrane, inhibiting the enzymatic activity, protein synthesis of RNA and DNA and others modifications that

can lead to apoptosis or necrosis of certain cells [9, 16–17].

Other studies indicate that the great amount of residual monomer released is in first 24 hours, after polymerization [8]. To minimize this great release throughout the time it is considered by some authors, a treatment after-polymerization, placing the dental prosthesis in water, during 24 hours or to place it in water at 50°C, during 60 minutes [5, 8–9]. Weaver and Garbel, in clinical practice, placed dental prosthesis in hot water and they verified that the cytotoxic effect on the patients was minor, because the great amount of free radicals released was during that period [8].

The objective of this study was to evaluate the amount of MMA released by a self-curing resin, in two types of solvent: distilled water and physiological serum. It was also evaluated the influence of temperature and time waited after preparation before the contact with the solvent.

2 EXPERIMENTAL PROCEDURE

2.1 *Materials and methods*

For this study 32 plates were manufactured in acrylic resin Lucitone Make Dentsply®, according to manufacturer indications: 20 g of powder (polymer) and 14 ml of liquid (monomer), polymerized at $40 \pm 3^\circ\text{C}$, at a pressure of 20–30 Psi, during 15 minutes. The plates had identical dimensions $45 \pm 0,1 \text{ mm} \times 2 \pm 0,1 \text{ mm}$, corresponding to the area of a base dental resin prosthesis. In order to evaluate the influence in the amount of PPM

released by three considered parameters: type of solution; incubation temperature; waiting time after manufacturing, the plates were divided into eight groups (Table 1).

The plates were individually placed in a volume of 20 ml of the respective solvent. For the spectrophotometric measurements an aliquot of 1,5 ml of solution was collected at different time points 30, 60, 120 and 240 minutes, after initial time. In order to keep the temperature of 37°C of four groups, the 150 mL beakers were used and samples were placed in a shaking -water bath covered with aluminum foil, to avoid evaporation. The ultra-violet absorption spectroscopy was used to quantify the amount of MMA in solution, for different considered solutions. First, it was determined a calibration curve according to the methodology suggested by Lamb et al, Stafford and Brooks. For that purpose, a calibration curve was prepared using pure monomer in five solutions with different concentrations (0, 1, 5, 10 and 50 ng/mL). The calibration curve is showed in the graph of Figure 1.

The measurements of absorbance were obtained using a wavelength of 210 nm, with a spectrophotometer Cintra 10e (Figure 2), because it is the wavelength for the absorption for MMA [18].

Data was analyzed using a SigmaStat 3.5 software and $p < 0,05$ was considered as of statistical level of significance. Analysis of results was performed using three-way ANOVA on ranks.

Table 1. Groups considered within this study and respective experimental conditions.

Groups	Solvent	T (°C)	Time (hours)	N
①	H ₂ O	20 ± 1	1	3
②	NaCl 0,9%	20 ± 1	1	3
③	H ₂ O	20 ± 1	24	3
④	NaCl 0,9%	20 ± 1	24	3
⑤	H ₂ O	37	1	5
⑥	NaCl 0,9%	37	1	5
⑦	H ₂ O	37	24	5
⑧	NaCl 0,9%	37	24	5

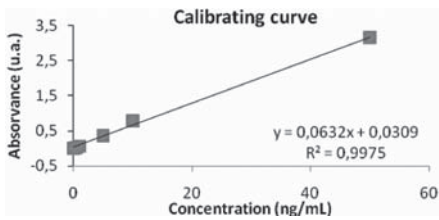


Figure 1. Calibrating curve, relating the amount of MMA and the respective absorbance.



Figure 2. General aspect of the spectrophotometer Cintra 10e, used within this study.

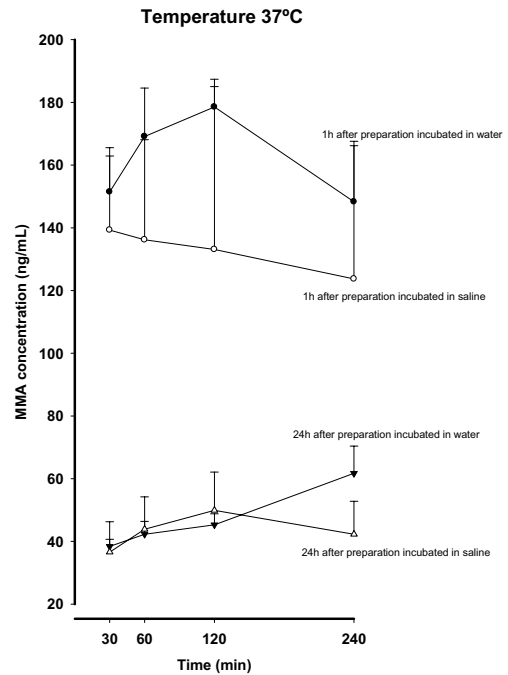


Figure 3. Release of MMA at 37°C to different solvents, after 1 hour or 24 hours preparation of the polymer. Data showed are mean $\pm$ SEM for 5 different experiments.

3 RESULTS AND DISCUSSION

All samples from each container, at different times, were analyzed and data grouped in Table 2.

The amount of MMA released to the solvent ranged from a minimum of around 6 ng/mL, observed 30 minutes after incubation to a maximal

Table 2. Methyl methacrylate (MMA) concentrations in solution expressed in ng/mL, obtained after incubation at different temperatures (room temperature – RT or 37°C) acrylate polymers plates prepared as described in materials and methods. Data was collected for 4 hours at different time points using two different solvents (water and NaCl 0,9%). Plates were prepared 1 hour or 24 hours before the experiments. Data is mean ± SEM for 3 to 5 experiments.

Experimental conditions			Time of contact with different solvents (minutes)			
			30	60	120	240
Temperature	Time	Solvent	MMA concentration (ng/mL)			
RT	1 h	H ₂ O	153,1 ± 43,4	261,8 ± 81,2	314,1 ± 91,4	533,1 ± 166,3
	1 h	NaCl	102,5 ± 33,8	162,3 ± 40,9	228,5 ± 9,9	303,7 ± 73,8
	24 h	H ₂ O	7,2 ± 0,7	17,0 ± 2,7	25,2 ± 2,5	37,5 ± 4,2
	24 h	NaCl	7,5 ± 0,6	13,5 ± 1,1	21,5 ± 1,9	31,9 ± 3,1
37°C	1 h	H ₂ O	151,4 ± 14,1	169,1 ± 15,5	178,5 ± 8,8	148,3 ± 17,9
	1 h	NaCl	139,3 ± 23,6	136,2 ± 31,9	133,1 ± 51,9	123,7 ± 43,9
	24 h	H ₂ O	38,4 ± 2,3	42,3 ± 4,1	45,3 ± 3,5	61,7 ± 8,7
	24 h	NaCl	36,6 ± 9,7	43,9 ± 10,3	49,9 ± 12,2	42,3 ± 10,5

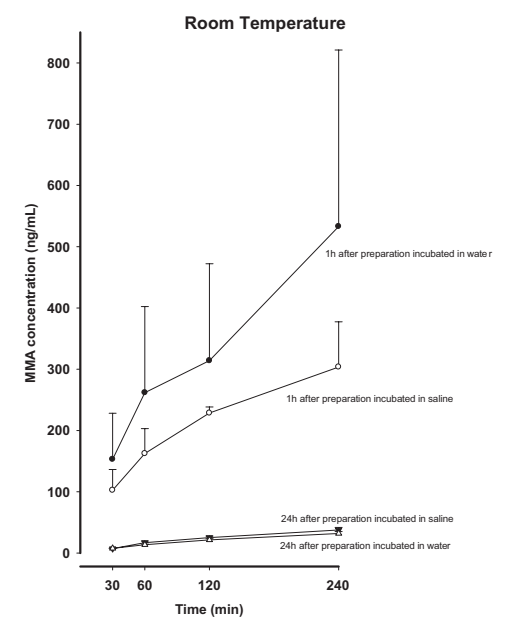


Figure 4. Release of MMA at room temperature to different solvents, after 1 hour or 24 hours preparation of the polymer. Data showed are mean ± SEM for 3 different experiments.

value of 775 ng/mL, observed 240 minutes after contact with the resin.

In this study, only the results for the first four hours are reported, because it was observed a reduction in the amount of MMA present in the solutions at 18 and 24 hours, probably due to its volatility and consequent evaporation.

There were no statistically differences for solvents tested at any given testing conditions.

When different temperatures were compared for the release of monomer, a difference was found but its statistical significance is marginal due to limited amount of data and large dispersion of results. Further experiments need to be carried out to increase the power of the statistical analysis. Despite this drawback regarding the power of the statistical analysis, a difference was observed for RT and 37°C for 24 hours after manufacturing of the base resin.

A statistical significant difference was found for all time points, when time of waiting after preparation was 1 hour compared to 24 hours (*p* < 0,001). Thus, the time elapsed after preparation of the base resins is a major determinant for the release of MMA.

4 CONCLUSION

With this study it was observed that the amount of MMA released is dependent to the time elapsed between preparation and usage. No others significant differences were observed. As others authors suggested, it is of great important to wait, at least, 24 hours, before using acrylic base dental resin prostheses.

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Muscular and articular forces exerted on the human mandible

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ABSTRACT: A joint prosthesis design takes into account data characterizing the natural biomechanics. To characterize the temporomandibular joint (TMJ) the approach had to be both kinematic (geometry, displacements) and quasi-static (forces, strains, stresses), the two being complementary. This paper presents a 3D method quantifying and analyzing the muscle and the joint forces exerted on the mandible under loadings. The experimental protocol, the hypothesis, the elaboration and the resolution of the equation system that traduces the mandible equilibrium, are described and then applied using the data of four volunteers. Dissections made it possible to describe the contacts between the mandible condyles and the temporal bone and as well the jaw closing muscle insertions. The directions of the muscular forces were thus determined in a morphological coordinate system. A sensor was designed to simulate the bite force between two teeth, successively between two incisors, two premolars and then two molars. MRI and electromyography results permitted, in vivo, to evaluate the magnitudes of six jaws closing muscle forces for each side of the face. The resolution of the equation system written to calculate the forces transmitted through the articular contacts was computer-assisted.

1 INTRODUCTION

Several TMJ total prostheses remained without significant diffusion. Persistence of pains or inflammatory phenomena pointed out the implant micro displacements [Karray, 99] [Layrolle, 06] [Baird, 98]. The rupture of the bone connection might happen when the load magnitudes were not perfectly taken into account [Speculand, 2000] [Wolford, 2003]. These problems that involve the knowledge of the loads, of the strains and/or of the stresses require a complementary data characterizing the TMJ biomechanics.

To characterize the healthy TMJ, an experimental protocol based on the functional analysis was elaborated [Mesnard, 05]. This approach pointed out the investigation directions and the experimental associated techniques. The articular displacements were determined using the 3D video analysis [Mesnard et al., 05]. The purpose is now to evaluate the muscular and the articular forces exerted on the mandible.

2 MATERIALS AND METHOD

2.1 Biomechanical model and volunteers

2.1.1 Reference coordinate system

The face was supposed to present a morphological symmetry with respect to the sagittal plane. The study used a functional reference system linked to the anatomical Camper's plane (Fig. 1). The coordinate system (S_c) was associated to the three morphologic points defining the plane: the sub nasal point, the two centres of the left and right condyles in the intercuspidation position. The origin C of S_c was the middle of the segment defined by the two condyle centres. The basis associated to these three axes was noted b .

2.1.2 Jaw closing muscles

Three pairs of muscles mainly rise up the mandible and contribute to the transmission of the forces that are useful for speaking or for chewing: the masseter muscles, the medial pterygoid muscles

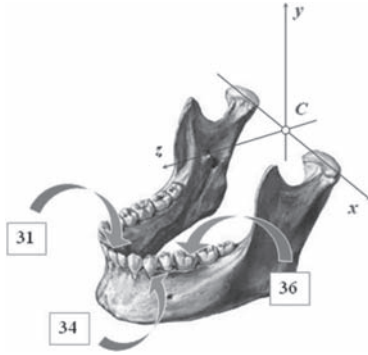


Figure 1. Biomechanical model.

and the temporal muscles. The masseter muscle, short and thick, presents two bundles. The surface masseter (*sm*), arises from the anterior part of the zygomatic arch. The deep masseter (*dm*), smaller, arises from the posterior part of the zygomatic arch. The medial pterygoid muscle (*mp*) remains parallel with the inner face of the mandible. The temporal muscle arises from the temporal fossa, passes through the zygomatic arch and is inserted onto the mandible coronoid process. The anterior bundle (*at*) simultaneously propels and lifts the mandible. The medial bundle (*mt*) whose fibers are quasi orthogonal to the occlusion plane primarily lifts the jawbone. Finally the fibers of the posterior bundle (*pt*) that also present some insertions on the TMJ meniscus act primarily in the movement of retro propelling.

2.1.3 Volunteers

The volunteers, medical students who had been duly informed and who had freely accepted the protocol, filed an assent form. They had undergone no ortho-dontic treatment likely to modify their articular morphological characters and muscular physiology.

2.2 Mandible static equilibrium

During an inter dental loading, the mandible is submitted to the actions of the gravity $\{Wm\}$, of the jaw opening muscles $\{Aopl/m\}$, of the jaw closing muscles $\{Acll/m\}$, of the external inter dental load $\{Ase/m\}$, and lastly, of the two elementary joints $\{Acm1\}$ and $\{Acm2\}$ (cranium acting on the left and right mandible condyles). So, the mandible static equilibrium can be written,

$$\{Wm\} + \{Aopl/m\} + \{Acll/m\} + \{Ase/m\} + \{Acm1\} + \{Acm2\} = \{0\}$$

If the gravity and the jaw opening muscles actions are not taken into account, the global

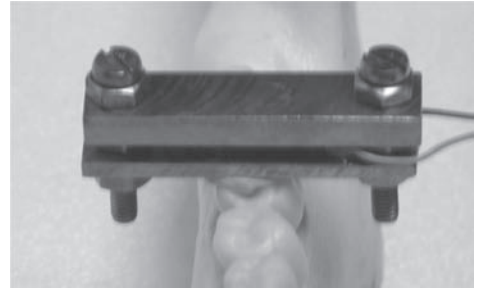


Figure 2. Inter dental sensor.

action, $\{Acm\}$, transmitted to the mandible can be written,

$$\{Acm\} = \{Acm1\} + \{Acm2\} = -\{Ase/m\} - \{Acll/m\} \quad (1)$$

Then it became necessary to quantify $\{Ase/m\}$ and $\{Acll/m\}$, the inter dental load force and the mandible closing muscle forces, to evaluate $\{Acm1\}$ and $\{Acm2\}$ the actions transmitted by the two joints.

2.3 Inter dental load

The static bite force that loaded a mandible tooth was simulated and recorded using an inter dental specific sensor (Fig. 2).

This sensor was also designed to allow controlling the mouth opening with two different thicknesses (5 and 15 mm). It was, in vivo, successively introduced between the two incisors (31–21), the two premolars (34–24) and then the two molars (36–26) to quantify $\{Ase/m\}$.

2.4 Directions of the jaw closing muscle forces and location of the points of contact

Four dissections made it possible to investigate the contacts between the condyles and the temporal bone and the insertions of the muscles. Photographs and an image processing software developed in the laboratory (Labview) carried out the experimental determination of the contact points and of the muscle insertion centres in the morphological system (Fig. 3). The calculation of the muscular force directions was thus computer-assisted for two different thicknesses of the sensor (5 and 15 mm). The co-ordinate values used to calculate the articular forces corresponded to the averages of the values recorded realizing the four dissections.

2.5 Magnitudes of the jaw closing muscle forces

In vivo electromyography and MRI data permitted to evaluate the magnitudes of six muscle forces.

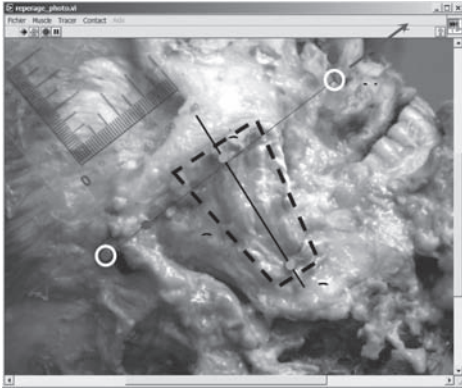


Figure 3. Determination of the muscle force directions.

In significant voluntary contractions, the global EMG informs about the activity of the muscle fibers and the magnitudes of the muscle forces. The maximum magnitude of these forces can be reached only when almost all the fibers are implied in the contraction process. The electric activity was here intercepted in an intramuscular bipolar mode. Intramuscular thread electrodes were used for the six muscles (platino-iridium, diameter 20–30 μm). As the two extremities of a muscle did not present any relative displacement, the muscle length remained constant and the contraction was isometric.

Under the isometric conditions described, a linear relation between the EMG effective signal and the muscular force exists. If the force developed by the muscle is noted F , the maximum value is then noted F_{max} . In the same way, $EMGr$ and $EMGr_{max}$ indicate the current and the maximum root mean square values (RMS) of the EMG signal. Then, the linearity can be traduced in the following relation:

$$F_{muscle} = (F_{muscle\ max}/EMGr_{max}) * EMGr,$$

where

$$EMGr_{RMS} = \left(\frac{1}{T} \int_{t-T/2}^{t+T/2} (MEG)^2 dt \right)^{1/2} (mV)$$

During the contraction, when almost all the muscle fibers are recruited (maximum bite force), there exists an equally linear relation between the maximum muscle force and its main section at rest. If the area of this section is noted S_{max} and if K indicates a constant coefficient that characterizes the muscular group, the relation can be written:

$$F_{muscle\ max} = S_{max} * K$$



Figure 4. Determination of the muscle EMG signals.

The values suggested for the coefficient K (N/m^2) vary with the muscle morphology and the experimental conditions. They were taken in the literature [Moller, 66] and the current muscular force was expressed by:

$$F_{muscle} = (S_{max} * K/EMGr_{max}) * EMGr$$

To use this relation, for each jaw closing muscle, it was thus necessary to determine $EMGr_{max}$, repeating isometric voluntary maximum contractions and S_{max} , starting from an MRI examination.

To exploit the EMG signals the device provided several successive functions: capture using thread electrodes (Fig. 4), amplification and filtering of the signal, recording and treatment of the data.

To reduce the strong embarrassment imposed to the volunteer, the morphological symmetry of the face was admitted. Consequently, the areas of the muscle sections were only red on the left profile. The electrodes were equally established on the left profile. The sensor was positioned successively on the incisors (31/41), the first premolars (34 then 44) and the first molars (36 then 46). Admitting a perfect geometrical symmetry of the face, the situation (31/41), corresponded to a plane loading. On the other hand, for a dissymmetrical loading, between premolars for example, the sensor was first of all positioned on the tooth 34. The electrodes and the sensor were then located on the same profile. The sensor was then moved on the tooth 44. The electrodes and the sensor were thus located on different profiles. This technique made it possible to record EMG signals of the two profiles avoiding a bilateral implantation of electrodes.

MRI provides images presenting a high degree of accuracy. To determine the main muscle section, the principal direction might be known. Several parallel and close recordings, orthogonal with this longitudinal muscle axis, were then realized.

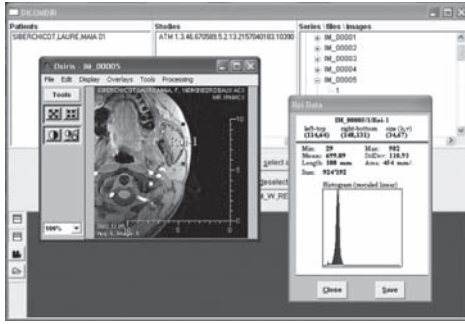


Figure 5. Determination of the jaw closing muscle sections.

The digitized images were analyzed using an image processing software (Osiris). A module makes it possible to select one zone of the image in order to characterize it. Thus, this area of interest can, as on the Figure 5, delimit the two masseters. The software revealed here a total area close to 454 mm².

2.6 Writing and resolution of the equation system

On the basis b , the action that can be transmitted by the left elementary joint was written:

$$\{A_{cm1}\} = \{X1\ x + Y1\ y + Z1\ z; L1\ x + M1\ y + N1\ z\} \text{ at point } C1,$$

where, the force was represented by, $X1\ x + Y1\ y + Z1\ z$ and the moment was represented at point $C1$ by, $L1\ x + M1\ y + N1\ z$. Traducing the mandible equilibrium at point C_2 , this moment was transformed as follow:

$$L1\ x + M1\ y + N1\ z + C2\ C1 \times (X1\ x + Y1\ y + Z1\ z) \text{ at point } C2.$$

On the basis b , the action that can be transmitted by the right elementary joint was written:

$$\{A_{cm2}\} = \{X2\ x + Y2\ y + Z2\ z; L2\ x + M2\ y + N2\ z\} \text{ at point } C2.$$

Without any hypothesis, the global action, $\{A_{cm}\}$, gathered twelve unknown components:

$X1, Y1, Z1, L1, M1, N1, X2, Y2, Z2, L2, M2$ and $N2$.

In the same way, the action exerted on the mandible by the muscle noted i , was written:

$$\{A_{muscle\ i/m}\} = \{F_i = F_i\ u\ i; O\} \text{ at point } B,$$

the insertion centre of the muscle i .

The writing was similar for the twelve muscles. The value of i varied from 1 to 12. For each muscle,

the algebraic effort F_i was calculated starting from the electromyographic data. The directions of the unit vectors u_i , corresponding successively to the two sensor thicknesses (5 or 15 mm) and the three positions of the sensor, were defined by the dissections. The components of u_i and, consequently, those of F_i on the basis b could thus be determined.

The global action of the jaw closing muscles gathered twelve known forces:

$$\{A_{cl/m}\} = \sum \{A_{muscle\ i/m}\} \text{ for } i \text{ varying from } 1 \text{ to } 12.$$

Lastly, the action corresponding to the load registered by the sensor was written:

$\{A_{se/m}\} = \{F = F\ u; O\}$ at point T , corresponding to the tooth 31, 34 or 36. The force F was recorded and thus known. The unit vector u was calculated, normal to the bisecting plane corresponding to the mouth opening. The components of u and those of F on the basis b can thus be calculated.

The hyper static system (1) traducing the mandible static equilibrium was written at point $C2$. The equations involved the twelve unknown components. If the laws that characterize the joint behaviors (meniscus comportment), the volunteer morphology (geometry) and the joint natures (degrees of freedom) are unknown, then the calculation of the forces transmitted under loading through the contacts remains impossible. The partial evaluation of these actions imposed more or less realistic assumptions that, in any rigor, should have to be validated.

Each diarthrosis involves a meniscus and a capsule. The presence of synovial fluid within the capsule makes rolling and sliding movements much easier. This short description made it possible to suppose that the four components $M1, N1, M2$ and $N2$ remain close to zero.

The muscle forces and the load being known, the two components $L1$ and $L2$ included the action of friction, the ligament actions and the action of the muscles that were not taken into account around the axis ($C2, x$). The equation of the moments around this axis should be verified; the partial resolution of the system thus used a numerical optimization that minimized a given criterion. The selected criterion consisted in minimizing the sum ($L1 + L2$). The six force components $X1, Y1, Z1, X2, Y2$ and $Z2$ remained unknown.

Starting from the geometrical analysis, two models of the TMJ were elaborated (Fig. 6). The first model (FM) corresponded to the symmetrical case of loading, case where the sensor occupied the position 31 (Fig. 6a). For each condyle, it comprised one point of contact or "ball-and-plane pair" (1BPP, joint of two links made up of a ball and a plane). The second model (SM) corresponded to the dissymmetrical case; the sensor

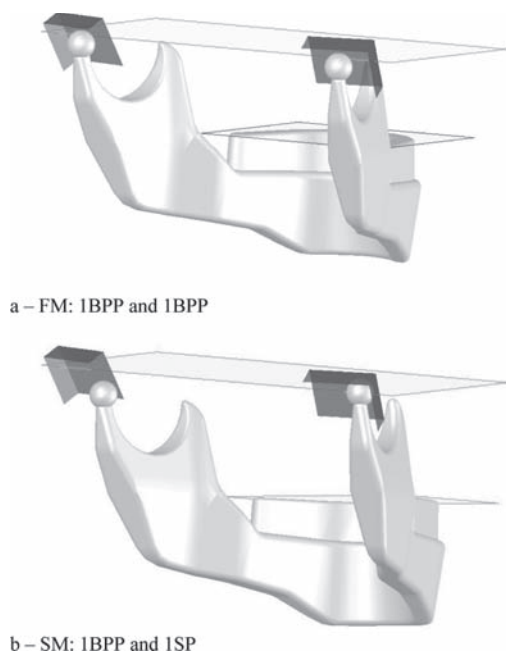


Figure 6. Geometrical models.

was then in the positions 34 or 36 (Fig. 6b). It presented one point of contact for the left mandible condyle or “ball-and-plane pair” (1BPP) and a “spherical pair” (1SP, joint of two links permitting spherical motion of one link relative to the other) for the right condyle.

To evaluate the force components, the resolution of the equation system was computer-assisted. For FM, a symmetrical loading, the components $X1$ and $X2$, opposed, remained unspecified. The two condyles were requested by equal sagittal components. For SM, a dissymmetrical loading, if the friction was neglected, therefore the $X1$ component was cancelled and $X2$ remained unspecified.

3 RESULTS AND DISCUSSION

3.1 Jaw closing muscle forces

The forces developed by the muscles in experimental situations could not be compared if the sensor did not indicate the same load for all the tested situations. To circumvent this difficulty, the muscular forces F_i were transformed in F_i unit for an inter dental force which value was always 1 N. The transformation was the following,

$F_i \text{ unit} = F_i / |F|$ where, F represented the force registered in the current test.

Table 1 presents the magnitudes of the forces for the six pairs of jaw closing muscles. Three tests

Table 1. Jaw closing muscle forces (N) corresponding to a load of 1 N on the tooth.

		Muscle					
		<i>dm</i>		<i>sm</i>		<i>mp</i>	
Sensor		Left	Right	Left	Right	Left	Right
31	5	4,72	4,72	18,72	18,72	20,78	20,78
34	5	3,65	4,65	8,16	11,87	8,62	7,05
36	5	2,99	3,78	7,26	7,32	13,59	10,41
31	15	6,02	6,02	21,73	21,73	42,16	42,16
34	15	3,53	3,54	5,89	7,72	20,36	8,76
36	15	1,67	2,80	3,75	4,90	4,86	3,68

		Muscle					
		<i>at</i>		<i>mt</i>		<i>pt</i>	
Sensor		Left	Right	Left	Right	Left	Right
31	5	1,07	1,07	0,51	0,51	0,35	0,35
34	5	5,07	1,73	2,71	0,52	0,65	0,33
36	5	6,32	4,51	3,13	1,70	0,03	0,18
31	15	3,70	3,70	2,59	2,59	0,43	0,43
34	15	4,72	2,47	2,19	1,04	0,77	0,64
36	15	3,29	2,71	1,50	1,63	1,02	0,17

Table 2. Jaw closing muscle forces (N) corresponding to a load of 1 N on the tooth.

compo.		$X1+X2$	$Y1$	$Y2$	$Z1$	$Z2$	$Z1/Y1$	$Z2/Y2$
31	5	0	-41,4	-41,4	6,0	6,0	-0,14	-0,14
34	5	0,3	-24,1	-24,9	4,2	0,9	-0,17	-0,03
36	5	1,2	-27,7	-25,7	4,7	3,1	-0,17	-0,12
31	15	0	-68,0	-68,0	4,5	4,5	-0,07	-0,07
34	15	5,1	-29,1	-24,3	5,1	3,4	-0,18	-0,14
36	15	0,3	-12,8	-14,3	4,1	2,8	-0,32	-0,20

were realised, in each position, for the four volunteers. The means are calculated starting from the twelve values.

The contribution of the posterior temporal muscle remains weak compared to that of the two other muscle bundles. The partition of the temporal muscle in two bundles seems sufficient. The objective had been here to increase the result precision and to circumvent the difficulty to vectorize this muscle that looks like a hand-held. Because of the volunteer hair, the use of external electrodes can be considered only for the anterior temporal. This reduction of the protocol is however interesting because it will simplify the implementation and will reduce the pain of the volunteer.

3.2 Symmetrical loading, FM model

The numerical values correspond to a force of 1 N recorded by the sensor (Table 2). The two mandi-

ble condyles are requested by equal sagittal components. For the mouth opening corresponding to the thickest sensor (15 mm), the articular points of contact are located near the temporal condyles. $Y1$ and $Y2$ reach then their maximum absolute value. The branch of the mandible, directed according to the y-axis by the opening, is then requested in quasi pure compression. The two components $Z1$ and $Z2$ remain small.

3.3 Dissymmetrical loading, SM model

For the mouth openings corresponding to the thinnest sensor (5 mm), $Y1$ and $Y2$ keep close values. From the position 31 to the position 36 the absolute values of $Y1$ and $Y2$ decrease significantly. The component $Z1$ remains very small and quasi-constant whereas $Z2$ tends towards zero. From the position 31 to the position 36 the absolute values of $Y1$ and $Y2$ also decrease using the thickest sensor (15 mm).

4 DISCUSSION

The knowledge of the muscular force directions that is necessary to traduce the mandible equilibrium could be considered in vivo. The development of a personalized 3D muscular model should be carried out using diffusion MRI. Diffusion MRI produces in vivo images of biological tissues weighted with the local characteristics of the water diffusion. The image-intensities are attenuated, depending on the muscle anisotropic microstructure in which the water molecules diffuse.

The directions of the forces exerted by the masseter and the medial pterygoid muscles remain quasi parallel. This remark makes it possible to consider the removal of the pterygoid electrode established in the oral cavity by assigning a corrected section to the masseter. Dividing the temporal muscle in two bundles, it will be interesting to evaluate the contributions of each bundle and to plan to consider only the anterior temporal. This approximation would partially compensate the fact of not taking into account the jaw opening muscle and would remove four electrodes.

The jaw closing muscular group selected was composed of six muscles and ignored the jaw opening muscles. The protocol described in this paper allows taking into account the jaw opening muscles.

ACKNOWLEDGEMENTS

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Stress distribution on cantilever dental prostheses

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ABSTRACT: According to the Glossary of Prosthodontics Terms, a fixed dental prosthesis is any dental prosthesis that is luted, screwed or mechanically attached or otherwise securely retained to teeth or dental implants that furnish its primary support. The shape of this kind of dental prostheses depends on the geometry of the tooth that it's replacing due to its multiplicity of concavities and convexities. The connector area between each tooth demands special requirements since it's the area with the highest stresses during mastication, with loads that may reach 500 N, or even 1000 N in mastication peaks. The stress values generated in these connectors are highest in a cantilever bridge, i.e. a bridge that is retained and supported only on one end by one or more abutment(s), when compared to a conventional dental bridge. Actually, to optimize the design of a fixed dental prostheses in a functional and esthetic rehabilitation, it's possible to combine engineering techniques and the new computer-aided-design/computer-aided-manufacturing systems (CAD-CAM) applied to Dental Medicine. In this work it is described a methodology to optimize the geometry and connection of a cantilever bridge in a fixed dental prosthesis, by means of experimental and numeric techniques, like photoelasticity and finite element meshes, respectively. In the work here described a CAD-CAM dental system (Kavo®Everest) was used to create a FDP with 4 teeth representing a conventional bridge and a cantilever bridge. The .igs file generated with this software had been converted to a Solidworks file in order to make a longitudinal cut of the structure. That design was augmented 10 times in order for being cutted on a photoelastic template and easily analyzed. The geometric model was also used to generate a FEM in the code ABAQUS. The obtained results are similar to those of other authors, like Anusavice and Oh. Highest stresses are verified in the gingival embrasure of the connector, specially the one related to the cantilever, while deformation occurs mainly in the occlusal embrasure. The introduction of a fillet on the connector area of the CAD design helps reducing the stresses generated in this area. The stress distribution in the experimental and numerical model was similar, and validated by Tresca criteria. We can conclude that fillet introduction is crucial to reduce the stresses in the connector area that may cause a catastrophic failure of the dental rehabilitation. CAD-CAM dental software should have this kind of design functions just like CAD-CAM engineering software.

1 INTRODUCTION

1.1 *Prosthodontics*

According to the Glossary of Prosthodontics Terms, (The Academy of Prosthodontics 2005) a fixed dental prosthesis is any dental prosthesis that is luted, screwed or mechanically attached or otherwise securely retained to teeth or dental implants that furnish its primary support. It may be a conventional bridge—retained on both ends—or a cantilever bridge—supported only on one end by one or more abutments.

1.2 *Biomechanics of dental prosthesis*

The shape of this kind of dental prostheses depends on the geometry of the tooth that it's

replacing due to its multiplicity of concavities and convexities. The connector area between each tooth demands special requirements since it's the area with the highest stresses during mastication, with loads that may reach 500 N, or even 1000 N in mastication peaks. (Fischer et al. 2004; Oh et al. 2002; Oh & Anusavice 2002; Okeson 2003; Pereira et al. 2006).

The stress values generated in these connectors are highest in a cantilever bridge.(Correia et al. 2009; Eraslan et al. 2005; Romeed et al. 2004).

Some authors studied the optimization of the cantilever connectors by introducing changes in its shape. (Correia et al. 2009; Oh et al. 2002) Previous studies (Correia et al. 2009) in a simplified model of a similar fixed partial denture framework revealed that the introduction of a fillet in

the gingival embrasure of the cantilever connector reduced stress distribution in the area.

1.3 Bioengineering

Actually, to optimize the design of a fixed dental prostheses in a functional and esthetic rehabilitation, it's possible to combine engineering techniques and the new computer-aided-design/computer-aided-manufacturing systems (CAD-CAM) applied to Dental Medicine. (Liu & Essig 2008).

In this work it is described a methodology to optimize the geometry and connection of a cantilever bridge in a fixed dental prosthesis, by means of experimental and numeric stress analysis techniques, like photoelasticity and finite element meshes, respectively.

1.4 Aim

To study stress distribution on a computer aided design of a cantilever fixed partial denture with different fillet sizes in the gingival embrasure of the cantilever connector, by means of experimental and numerical stress analysis techniques.

2 MATERIAL AND METHODS

In the work here described a CAD-CAM dental system Everest ([®]Kavo Dental GmbH) was used to

create a fixed partial denture titanium framework with 4 teeth representing a conventional bridge and a cantilever bridge involving the canine, first and second premolar and first molar (ex. teeth 33–36), supported by two dental implants (ex. teeth 33 and 35).

The connectors between all teeth had a 5,28 mm² section area.

The .igs file generated with this software had been converted to a Solidworks v.2007 ([®]Daussalt Systems S.A.) file and the design was optimized to correct some intersection defects from the original Everest CAD image.

This geometric model was used to generate a finite element mesh in the code ABAQUS ([®]Daussalt Systems S.A.), where a stress analysis was performed with a load application on the occlusal table of the cantilever tooth (1 N) and different size fillets (0; 0,5 and 1 mm) in the gingival embrasure of the cantilever connector.

Then, a longitudinal cut of the CAD image with the 1 mm fillet was performed.

That cantilever shape was augmented 10 times in order to being cutted on a photoelastic template and easily analyzed on a transmission polariscope (Vishay[™]).

Stress analysis was first performed on the cantilever tooth with a 1 mm fillet, followed by 0,5 mm and 0 mm fillets. Load was applied on different points of the cantilever tooth: near the connector, in the middle of the tooth and in the distal crest of the tooth.



Figure 1. Computer-aided-design of a fixed partial denture with a cantilever molar tooth.



Figure 2. CAD image obtained from Solidworks.

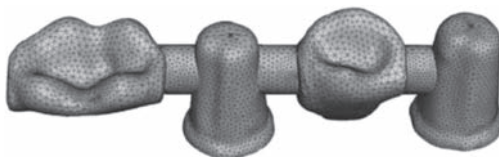


Figure 3. Finite element mesh of the CAD image.

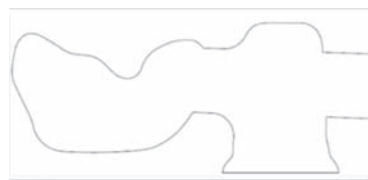


Figure 4. Longitudinal cut of the CAD image.

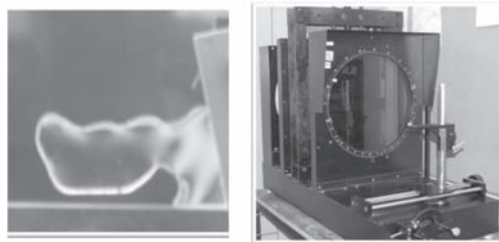


Figure 5. Photoelastic model (left) to be analysed on a transmission polariscope (right).

3 RESULTS

Stress analysis on the finite element mesh showed that the highest stresses where localized in the gingival embrasure of the cantilever tooth (Figure 6).

Different size fillets (0–0,5 and 1 mm) were introduced in this gingival embrasure. Results from the finite element stress analysis are exposed on Figure 7 and Table 1. The introduction of the fillet resulted on a reduction of the maximum von Mises stress levels.



Figure 6. Finite element mesh of the cantilever tooth showing highest stresses in the gingival embrasure.



Figure 7. Finite element mesh of the cantilever tooth with different fillets on the connectors (left: 0 mm; middle: 0,5 mm; right: 1 mm).

Table 1. Von mises maximum stress values obtained in the different FEM analysis.

0 mm	0,5 mm	1 mm
4,996 MPa	3,810 MPa	3,189 MPa

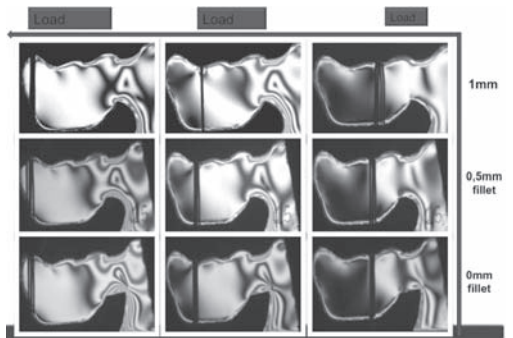


Figure 8. Photoelastic model with different loads and fillets.

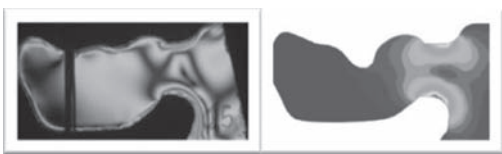


Figure 9. Tresca criteria for stress analysis validation of photoelastic model.

An experimental stress analysis was also performed with the photoelasticity technique. The results obtained from the photoelastic stress analysis can be visualized on the following figure.

Stress fringe patterns are more concentrated in the gingival embrasure of the cantilever tooth. The distal load point application showed patterns with more fringes then the point near to the connector.

The obtained results are similar to those of other authors (Correia et al. 2009; Eraslan et al. 2005; Oh et al. 2002; Oh & Anusavice 2002). Highest stresses are verified in the gingival embrasure of the connector, specially the one related to the cantilever, while strains occur mainly in the occlusal embrasure. The introduction of a fillet on the connector area of the CAD design helps reducing the stresses generated in this area. The stress distribution in the experimental and numerical model was similar, and validated by Tresca criteria (Figure 9).

4 CONCLUSIONS

From this work can be concluded that fillet introduction is crucial to reduce the stresses in the connector area that may cause a catastrophic failure of the dental rehabilitation. CAD-CAM dental software should have this kind of design functions just like CAD-CAM engineering software.

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Clinical evaluation of implant-retained overdentures: Mechanical complications

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ABSTRACT: Implant retained overdentures is an effective method of treatment for edentulous patients when unstable prosthesis and residual ridge resorption are present. The usual complications mostly consist of screw loosening, gingival hyperplasia, and needs for replacement of clips, denture relining and readjustment of occlusion and articulation. This study aims to report the clinical mechanical complications and to suggest possible relations with functional, mechanical and technical aspects. Twenty-one files of patients treated with removable acrylic implant-retained prosthesis using clip-bar attachment system were evaluated. The 3 major mechanical complications found, in mean incidence per patient/year, were: failure of the attachments (1.24); occlusal dissadaptation (0.42) and prosthesis damages (0.41). The mechanical complications are a common important event and close follow-up is inevitable. More studies are needed in order to make possible an integration of all relevant concepts, and enable a planning based on evidence, more correct and individualized.

1 INTRODUCTION

The use of overdentures retained by implants, often in reduced number, 2 to 4 implants, is an effective method of treatment, safe and with good predictability to edentulous patients, especially when unstable removable prostheses and mild to severe residual ridge resorption are present (Assunção 2008, Karabuda 2008). It offers a solution with a high longevity, great acceptance and satisfaction, preserving the residual alveolar ridge, improving retention, stability and chewing efficiency (Assunção 2008). There are described and commercially available, various connecting systems, which can generally be divided into “splinting”, the bar and docking systems or “not splinting”, with attachments, being the most common the ball type, magnets, or use of telescopic copings. The “Splinting bar-clip” system, although expensive and technically demanding, is one of the most used and reliable, avoiding difficult and unpredictable surgical interventions (Karabuda 2008). Despite complications are usual, they are frequently not serious, consisting in loosening of screws, gingival hyperplasia, and need for replacement of clips, denture relining and readjustment of occlusion and articulation (Geertman 1996).

2 AIM

The principal aims of this study are to report the mechanical complications that arise in the

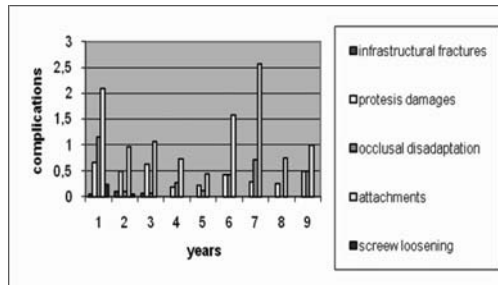
clinical practice and to suggest possible relations to biomechanical and technical aspects of implant retained removable prostheses and functional intervenients.

3 MATERIAL AND METHODS

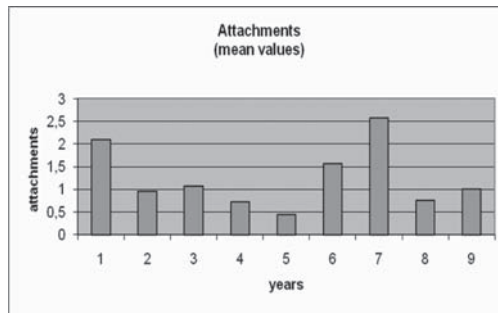
There were evaluated, retrospectively, the clinical and panoramic radiology files of 21 patients, who were treated for oral rehabilitation in a private practice dental office in Oporto, Portugal, with removable acrylic implant-retained prosthesis using clip-bar attachment system, supported by different number of implants with different conformations of the bar, height and thickness. The selection criteria was to include all patients treated this way, who had scheduled appointments between October 1st, 2008 and February 1st, 2009. There were radiologically evaluated the antegonial notch as a sign for muscular hyperactivity (coded 0 for flat and 1 for prominent), as well as bone loss around implants (coded 0 for absence of vertical signs of bone loss, 1 for reduction inferior to 1/3 of implant length, coded 2 for 1/3 to 1/2 of the length and coded 3 when superior to 1/2 of implant length). The type of opposing dentition, new prosthesis and the biological and mechanical complications that these rehabilitations presented over the time were pointed, and evaluated clinically or radiologically. Statistical treatment was performed using Microsoft Office Excel 2003 and conventional statistical formulas.

4 RESULTS

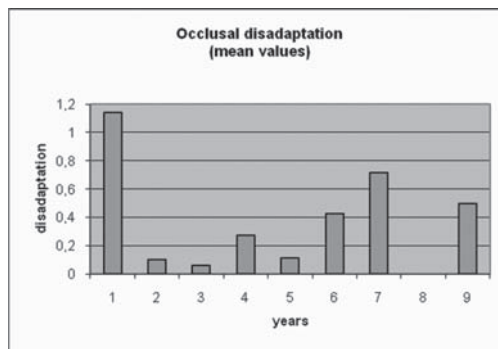
The technical and mechanical complications are higher for overdentures than for other implant solutions (Berglundh 2002). The mechanical complications used for evaluation purposes were infrastructural fractures, prosthesis damages, occlusal dissadaptation, failure of the attachments and screw loosening. The 3 major mechanical complications found, in mean incidence per year, were: failure of the attachments (1.24); occlusal dissadaptation



Graphic 1. Mechanical complications (mean values).



Graphic 2. Replacement of attachments by patient per year.



Graphic 3. Occlusal dissadaptation by patient per year.

(0.42) and prosthesis damages (0.41) (Graphics 1 to 3). Also two in twenty-one patients had the need for a new prosthesis in the first year of ware. Seven in twenty-one patients had the need for new prosthesis along the time. Four events of infrastructural fractures had been described. Logistic regression model was not conclusive to establish any relation between antegonial notch (sign for muscular hyperactivity), mechanical complications and bone loss.

5 DISCUSSION

Despite some disagreements there is an established idea that implant retained overdenture should be the first choice of care for edentulous individuals (Assunção 2008, Karabuda 2008). The Bar & Metal clip attachment system is considered to present medium retention characteristics (Chung 2004). In the literature available, prosthetic complications are the most referenced in removable implant-supported rehabilitations. Problems with the prosthesis and the infrastructure are common and important situations, often compromising the patient satisfaction and durability of the work, putting at risk the viability of the treatment plans and possible leading to loss of implants, or inability to use them in further prosthetic rehabilitation. Though they may be being ignored biological implications over the arches and the supporting tissue around the implants, particularly in extreme circumstances or in cases of deficient surgical planning/execution. In any case, the biomechanics is a critical factor in all stages, from the planning and treatment to the follow up.

The excess of functional loads generates stresses that are dissipated from the retention system to the implants and supporting tissue, and the intensity and amplitude of bone resorption is determined by the transmission and distribution mechanism of each retention system (Assunção 2008).

In the present study the failure of the attachments (1.24) was the most frequent problem, probably due to this component fragility and the use of a mouth-direct technique for their insertion. Infrastructural fractures, prosthesis damages, occlusal dissadaptation, failure of the attachments and screw loosening are all relatively frequent complications and one should expect them when using this type of implant rehabilitation. These results are difficult to compare, but they meet the literature expectations. It was impossible to obtain any correlation (logit model) involving antegonial notch (as marker for muscular hyperactivity) and mechanical complications. In somehow, usual clinical judgments, expect to experience more complications in patients with muscular hyperactivity. Seven in twenty-one patients had the need for new prosthesis along the time, probably due to dissadaptation or aging

of the prosthesis. In one situation, along eight years three new prostheses had been made likely due to the small free space reserved to the prosthetic rehabilitation which lead to precocity of major signs of wear and disrupt. It is important to note that happened four infrastructural fractures. The outstanding 2002 literature review paper by Berglundh et al. reported in their meta-analysis a 5.86% of implant loss during function, being the most frequent the maxillary ones, and twice as higher than in fixed reconstructions. They also described a 0.38% of implant fracture. It was obtained a 5 year incidence per patient mean complications of 1.56 for implant connection components and 1.90 for suprastructure. The incidence complications related to implant components and connections are higher than did in implant fixed reconstruction, and in suprastructure this incidence is 4 to 10 times higher. The implant fracture is an uncommon situation in overdenture rehabilitations (Berglundh 2002).

The first year is critical, since most of the adaptations and the larger number of problems takes place (Geertman 1996). In any way as concluded in the study from den Dunnen et al., the results from the present evaluation confirm the need for close routine follow-up services of hygiene care, adjustments, and treatment of complications (den Dunnen 1997). Quoting the work of Gantz in 1997, from the biomechanical point of view, the prognosis for overdenture treatments in patients wearing osseointegrated implants is maximized when: the number is as large as possible; the bar system used has been designed highest possible rigidity and the overdenture itself has been designed to allow some functional flexibility (Gantz 1997). However Klemetti in 2008 points that if a treatment decision is taken in favor of a removable overdenture, there still remains the question whether to fabricate a complete implant-supported overdenture on at least four implants, or an implant-retained and soft-tissue supported overdenture on two implants (Klemetti 2008). The use of implanted retained/supported overdentures is a very good choice in the treatment of edentulous persons and one should adapt the attachment system to personal characteristics, for instance, elderly patients with limited motor ability or dependent patients would probably need less retentive attachments (Chung 2004). In any case, the biomechanics is a critical factor in all of these stages: planning, treatment and follow up. It is crucial to attend to the biomechanical concepts, including face type, force of mastication, the axis and distribution of forces, the type of occlusion, the construction materials of infra and suprastructures, as well as the types of plug-in, the support tissues, the number and distribution of implants and the antagonist tooth quality in order to be possible to ensure a predictable and sustainable rehabilitation.

6 CONCLUSION

In implant retained or supported overdentures close follow-up is of the most relevant importance. The need for regular recall is indispensable for adequate preservation and stomatognathic function evaluation. Mechanical complications are usual and widely described. Both clinicians and patients should be aware of specificity of these rehabilitations and should accept the standard complications as normal event. Thus it is consensually accepted that all these rehabilitations require frequent maintenance, being the most important problems disclosed in first year of use. It's imperious to fully understand all the mechanical aspects and their relation with biology and functional occlusion, and science should keep the quest for optimization of oral rehabilitation in edentulous patients, usually elderly or compromised. Understanding the biomechanics of this type of rehabilitation, should be considered the key to clinical success. Both mechanical and biological factors intrinsic to the patient, or to the materials, the design of the infra and suprastructure and the distribution of the implants, should be observed from the point of view of its performance. More studies are needed in order to make possible an integration of all relevant concepts, and enable a planning based on evidence, more correct and individualized.

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Bioactive nanohybrid scaffolds mimicking natural dentin xenotransplanted in immunodeficient mice

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ABSTRACT: Nanohybrid scaffolds of poly(ethyl methacrylate-*co*-hydroxyethyl acrylate), P(EMA-*co*-HEA), pure and with 15 wt% of silica, with aligned tubular pores were prepared with the aim of mimicking structure and properties of the mineralized tissue of natural dentin. Half of the scaffolds with 15 wt% SiO₂ were hydroxyapatite, HAp-coated by immersion in simulated body fluid, SBF, for 14 days. Their physico-chemical and mechanical properties were investigated. The different scaffolds were implanted subcutaneously into immunocompromised nude mice for 4, 6 and 8 weeks and their biological response was analyzed. The HAp-coated scaffolds presented the best cellular distribution and neo-dentinal pattern, and neoangiogenesis. These tubular porous structures can be a useful tool for caries treatment, since they resemble natural dentin with regard to its structure and properties, induce deposition of HAp on their surfaces, through which they will integrate in the host mineralized tissue, and are colonized by cells, showing immunopattern and ultrastructural differentiations imitating the dentin structure.

1 INTRODUCTION

In the last years, different Tissue Engineering therapies to restore partial dental structures have been proposed [Bohl *et al.* 1998, Buurma *et al.* 1999, Young *et al.* 2002, Dobie *et al.* 2002, Nakashima 2005, Yelick & Vacanti 2006, Murray *et al.* 2007], based on the regenerative capacity of dental/pulp tissue, alternative to the conventional irreversible endodontic treatments. One of these therapies aims at inducing the regeneration of tooth structures by means of (cell)/scaffold constructs where the prefabricated natural or synthetic scaffold works as extracellular matrix for adhesion, proliferation and differentiation of cells [Nakashima 2005, Murray *et al.* 2007]. In the line of these strategies, the purpose of this work was to prepare silica-based organic-inorganic nanocomposite scaffolds mimicking the structure and properties of the mineralised matrix tissue of natural dentin, and investigate the

possible cellular colonization. Silica produces a reinforcing effect in the polymeric matrix and confers bioactivity to the surface of the scaffolds. The obtained porous structures are expected to stimulate the invasion by odontoblast prolongations of the pores of the scaffold when implanted *in vivo*, which would be able to fabricate new natural dentin while providing a very good adhesion between the scaffold and the surrounding natural tissue.

2 MATERIALS AND METHODS

2.1 Preparation of the scaffolds

Tubular porous scaffolds of P(EMA-*co*-HEA)/SiO₂ with 0 and 15 wt% SiO₂ were obtained by copolymerization of ethyl methacrylate, EMA (99%, Aldrich), and hydroxyethyl acrylate, HEA (96%, Aldrich), with 70/30 wt% of monomers ratio and

simultaneous acid-catalyzed sol-gel polymerization of tetraethoxysilane, TEOS (98%, Aldrich), in the presence of a fiber template.

Firstly, both organic monomers were mixed together with a 0.5% monomer weight of ethylene glycol dimethacrylate, EGDMA (98%, Aldrich), as crosslinking agent and a 2% monomer weight of benzoyl peroxide, BPO (97%, Fluka), as thermal initiator. Separately, a solution was prepared by mixing TEOS with distilled water and hydrochloric acid (37%, Aldrich) in the molar ratio 1:2:0.0185, respectively. The silica content was changed from 0 (hereafter H00 scaffolds) to 15 wt% (hereafter H15 scaffolds) by controlling the (EMA+HEA)/TEOS ratio and assuming that the sol-gel reactions were complete.

Glass tubes of 3 mm inner diameter were cut in 3.5 cm long pieces, stuffed with 10 μ m diameter polyacrylonitrile (PAN, Montefibre), aligned fibers and sealed on one side. The previously obtained mixtures were vacuum-injected in these glass moulds, which were immediately capped and placed in an oven for polymerization at 60°C for 21 h followed by post-polymerization at 90°C for 18 h. Afterwards, the PAN fibers were eliminated from the materials by dissolution in N,N-dimethylformamide (99.8%, Aldrich). The resulting scaffolds were washed in ethanol to remove residuals and unreacted monomers and finally dried in a vacuum desiccator at 80°C until constant weight.

Therefore, scaffolds of poly (ethyl methacrylate-co-hydroxyethyl acrylate), P(EMA-co-HEA) (H00) and P(EMA-co-HEA) with 15 wt% of SiO₂ (H15) with aligned tubular pores were obtained.

2.2 Hydroxyapatite (HAP) coating

The ability of the H15 scaffolds to form hydroxyapatite, HAP, on their surfaces was tested *in vitro*, by soaking them for 14 days in a simulated body fluid solution, SBF, with ion concentrations nearly equal to those of the human blood plasma, by the method proposed by Kokubo [Abe *et al.* 1990, Kokubo & Takadama 2006]. In order to obtain the SBF, two solutions were prepared. Solution 1 consisted in 1.599 g of NaCl (99%, Scharlau), 0.045 g of KCl (99%, Scharlau), 0.110 g of CaCl₂·6H₂O (99%, Fluka), and 0.061 g of MgCl₂·6H₂O (Fluka) in deionized ultra-pure water (Scharlau) up to 100 ml. Solution 2 was prepared by dissolving 0.032 g of Na₂SO₄·10H₂O (Fluka), 0.071 g of NaHCO₃ (Fluka), and 0.046 g of K₂HPO₄·3H₂O (99%, Aldrich) in water up to 100 ml. Both solutions were buffered at pH 7.4, by adding the necessary amounts of aqueous 1 M tris-hydroxymethyl aminomethane, (CH₂OH)₃CN (Aldrich), and 1 M hydrochloric acid, HCl, (37%, Aldrich). Then, both solutions were mixed to obtain SBF with the

corresponding molar ion concentrations: 142 Na⁺, 5.0 K⁺, 1.5 Mg²⁺, 2.5 Ca²⁺, 148.8 Cl⁻, 4.2 HCO₃³⁻, 1.0 HPO₄²⁻, 0.5 SO₄²⁻ mM.

Pieces of scaffolds of 3 mm length were suspended from a cotton thread inside closed glass vials filled with SBF. The ratio of geometric surface area of scaffold (without considering the inner pores surface) to solution volume was 0.12 mlmm⁻². The SBF solution was not renewed during the first 7 days. After the 7th day, a 2xSBF solution (with ion concentrations adjusted to twice those of SBF) with renovation was employed. This solution was renewed each 2–3 days, in order to provide more favourable conditions for apatite deposition [Tanahashi *et al.* 1994, Kim *et al.* 1999, Oliveira *et al.* 2003].

Samples were withdrawn from the SBF after 14 days, rinsed with water, room conditioned and finally dried in a vacuum desiccator at 80°C. Hereafter these scaffold will be referred to as H15HAP.

2.3 Characterization of the scaffolds

Morphological analysis of the scaffolds was undertaken by Scanning Electron Microscopy (SEM), in a JSM-6300 microscope, with the samples previously sputter-coated with gold, 15 kV of acceleration voltage and 15 mm of working distance. SEM images were obtained in both longitudinal and transversal sections.

The quantification of silica and the compositional changes occurring during the test in SBF were elucidated by microanalysis in and Energy Dispersive X-ray Spectrometer, EDS, attached to the Scanning Electron Microscope. Samples were previously sputter-coated with carbon under vacuum. Spectra were taken at 10 kV of acceleration voltage and 15 mm of distance working. Silicon was employed as optimization standard.

The porosity of the scaffolds was calculated afterwards through the specific volume of the respective bulk sample, obtained elsewhere [Vallés *et al.* 2009], and the weight and total volume (calculated geometrically) of the scaffolds, in duplicate. The following equation was employed:

$$\pi = \frac{V_{\text{pores}}}{V} = \left(1 - \frac{m_{\text{scaff}} \cdot v}{V} \right) \cdot 100$$

where V_{pores} is the pores volume, V is the overall sample volume, v is the specific volume of the corresponding bulk sample, and m_{scaff} is the weight of the scaffold.

The swelling of the samples in water and in SBF was quantified by weighing the samples dried and after equilibration to constant weight immersed in distilled water or in SBF (only 48 h in order to avoid HAP deposition) at room temperature using

a Mettler AE 240 balance with a sensitivity of 0.01 mg. The equilibrium contents were defined as the ratio between the mass of water or SBF in the sample and the mass of the dry sample.

Mechanical compression assays were performed on the scaffolds in the longitudinal direction (the tubular pores in the vertical direction, which will be their position when the scaffold is implanted in the tooth) in a Seiko TMA/SS6000 equipment, from 0.05 to 150 g, at 10 g/min, at room temperature. Measurements were repeated five times for each composition. Samples were cylindrical, 3 mm in diameter and approximately 3 mm in height.

2.4 Subcutaneous implants in vivo

The obtained scaffolds of P(EMA-co-HEA) (H00), P(EMA-co-HEA)/15 wt% SiO₂ (H15) and P(EMA-co-HEA)/15 wt% SiO₂ with the tubular surface coated with HAp (H15HAp) were implanted subcutaneously into the backs of ten 4 weeks old Balb/c male nude mice (IFA-CREDO, Lyon, France). Half of the scaffolds were immersed in a fibronectin solution (0.1% solution from bovine plasma, Sigma) overnight before the implantation. Mice were sacrificed after 4, 6 and 8 weeks.

Material for optical study was fixed in buffered formalin, paraffin embedded and stained with hematoxylin-eosin. The specimens for ultrastructural study, immediately after extraction, were fixed in 2.5% glutaraldehyde (buffered pH 7.4 in Sorensen phosphate solution). Next the tissue was post-fixed for 2 hours in 1% osmium tetroxide in buffer, and after dehydration with graded concentrations of acetone, the tissue blocks were embedded in Epon 812 (TAAB Lab, England). Semi-thin Epon sections (1 µm) were stained with toluidine blue for control in a light microscope and trimmed for ultrastructural study. Ultra-thin sections were cut with a diamond knife using a Reichert ultramicrotome (Leica, Illinois), contrasted with uranyl acetate and lead citrate, and examined in a Jeol Jem 1010 (Tokyo, Japan) electron microscope operated at 60 kV.

3 RESULTS AND DISCUSSION

3.1 Characterization of the scaffolds

Figure 1 displays the scanning electron microscopy images of the scaffolds. The transversal sections of the scaffolds of H00 and H15 show in all cases a homogeneous distribution of cylindrical pores of approximately 8 µm of diameter. The synthetic scaffolds resemble very much natural dentin with regard to the structure and distribution of the pores, being the diameter of the pores still large. Neither the pores diameter nor their distribution

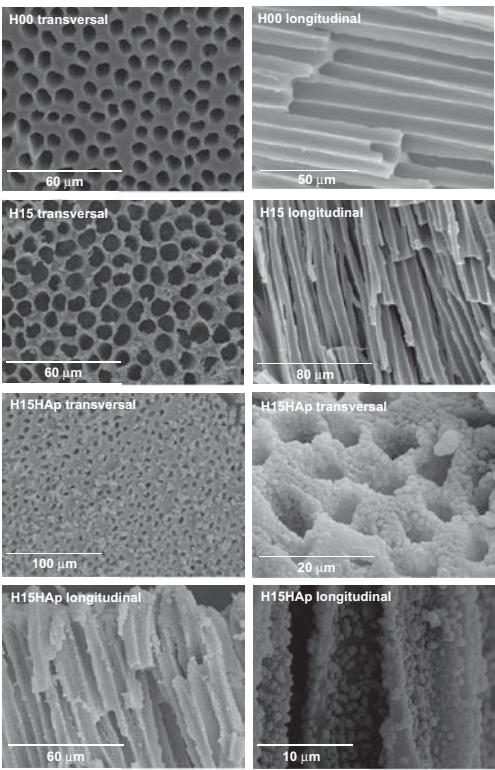


Figure 1. P(EMA-co-HEA) scaffold (H00) in the transversal and longitudinal section; P(EMA-co-HEA)/15 wt% SiO₂ scaffold (H15) in the transversal and longitudinal section; HAp-coated P(EMA-co-HEA)/15 wt% SiO₂ scaffolds (H15HAp) in the transversal and longitudinal section.

seem to vary with the composition. The porosity of H00 scaffolds is 37.26% and that of H15 scaffolds increases to 52.25%, but it can be modulated by employing different amounts of porogen fibers inside the glass tube.

After 14 days in SBF (H15HAp), the exterior transversal surface is covered with needle-shaped crystals forming typical cauliflower HAp structures [Rhee *et al.* 2002, Rhee 2003, Oliveira *et al.* 2003, Costa *et al.* 2003, Costa *et al.* 2005, Balas *et al.* 2006, Kokubo & Takadama 2006, Kawai *et al.* 2007], with an average diameter of 1 µm, which do not clog up the pores cavities. As a consequence, the average tubule diameter decreases from 8 to approximately 5 µm. This new pore diameter resembles more that of the natural dentinal tubule, being thus more convenient for the proposed application. Scattered aggregates of cauliflowers (with larger diameter) of what seem to be subsequent layers are superposed to the first homogeneous coating. The longitudinal fracture shows that the

surface of the tubular pores is almost completely coated with the first continuous apatite layer after 14 days in SBF in the regions near the transversal surfaces, but somewhat less in the inner zones.

The SiO_2 experimental contents (calculated from the Si average contents) and the amounts of the different elements deposited on the surfaces from the SBF were quantified by EDS. Figure 2 shows the EDS profiles of H15 and H15HAp scaffolds. The silica experimental contents in H15 are 15.33% in the bulk and 15.09% at the surface, *i.e.* they correlate very well with the nominal silica content. In the case of H15HAp, Si was not observed, Ca and P are the main elements, and other elements like Na and Cl were also detected. The average Ca/P ratio was 1.56, slightly lower than that of physiological HAp Ca/P ratio, which is 1.65 [Kim *et al.* 1999].

The equilibrium water content of the H00 scaffolds was 8.34%, and that of the H15 scaffolds was 6.92%. The samples were also swollen in SBF and the equilibrium contents were 5.60% and 7.02% for H00 and H15, respectively.

The compressive elastic moduli of the scaffolds were obtained from the slope of the stress-strain curves performed in the longitudinal direction. The elastic modulus of the H00 scaffolds was 3685 MPa and that of H15 increased to 6156 MPa. The compressive elastic modulus of the H15HAp scaffolds was 8192 MPa, 1.33-fold that of the original H15 scaffolds, but still lower than that of human dentin, 14.47 GPa [Craig & Peyton, 1958]. When implanted *in vivo*, the regenerated mineralized tissue is expected to contribute to the definitive compressive modulus.

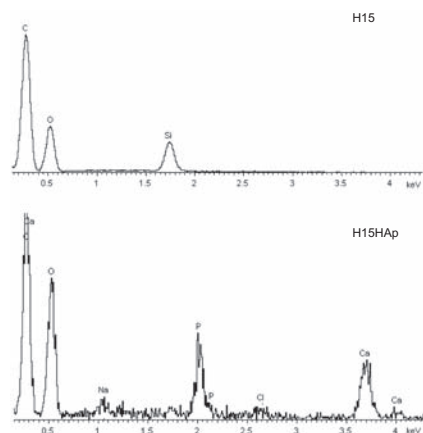


Figure 2. EDS spectra of the P(EMA-*co*-HEA)/15 wt% SiO_2 scaffold (H15) and the HAp-coated P(EMA-*co*-HEA)/15 wt% SiO_2 scaffold (H15HAp).

The HAp-coating of the pores and the consequent decrease in tubules diameter and mechanical reinforcement are very interesting features in the way to mimic natural dentin with regard to its structure and mechanical properties.

3.2 Biological response

Figure 3 shows some histologies as examples. The histologies of the explants of H00 and H15 scaffolds presented a good cellular distribution and intense neoangiogenesis, which was improved by fibronectin. Nevertheless, H15HAp scaffolds presented the best cellular distribution and neo-dentinal pattern. In this case fibronectin did not modify the results. Cellular nesting decreased after 4 weeks in H00 and H15 scaffolds, whereas it was stable until 8 weeks in H15HAp samples.

The cellular typification was carried out by transmission electron microscopy. It was observed that that the cellular colonizing response of the scaffolds changed according to the number of elements that

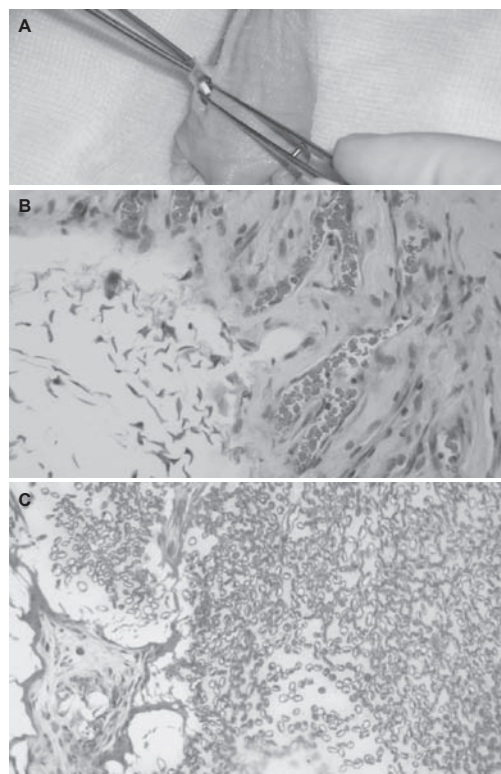


Figure 3. (A) Subcutaneous implantation of a scaffold in nude mice, (B) histology of the H00 scaffold (with fibronectin) after 4 weeks, (C) histology of the H15HAp scaffold after 4 weeks.

invaded the dentin neo-matrix, but this process was independent of the cellular typology.

The H00 scaffolds showed a tridimensional pattern with reticular appearance. Cellular characterization showed cells with immature fibroblastic characteristics (vs. mesenchymal). The reticular fibres are formed by slim collagen III fibres associated to the fibronectin, which is characteristic of pulpar tissue.

At ultrastructural level, the H15 and the H15HAp scaffolds displayed a mimetization of dentinal pattern, imitating in some cases a reactive dentin, because the cellular prolongations were included and also the cytoplasm and nucleus. In the non-mineralized scaffolds although the nested cellularity of central location of the scaffolds showed a cytoplasmic differentiation with a matrix secretory character, a high quantity of collagen surrounding the cellular net was evident. In the mineralized scaffolds, with narrower tubules, the collagen quantity surrounding the nested cells was lower due to the close adhesion to the walls, but it was clearly its secretor cell character.

Some cells in the surrounding areas presented lisosomes allowing their typification as reactive elements.

4 CONCLUSIONS

It can be concluded that the use of these bioactive nanohybrid scaffolds can be a promising tool for caries treatment since they are colonized by cells, showing immunopattern and ultrastructural differentiations imitating the dentin structure. Colonization and viability are improved by the use of mineralized interphases. Besides, the HAp-coating reinforces mechanically the scaffolds, decrease their pores diameter to a value closer to that of the dentinal tubules, and will likely facilitate the integration in the host mineralized tissue.

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Manufacturing and evaluation of the effectiveness of a new custom mouthpiece for wind instrumentists

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ABSTRACT: Playing a wind instrument requires a complex neuromuscular activity. The musician has to keep under effective control the column of expiratory air passing from the player into his instrument. In an orchestra there are different kinds of wind instruments that require a unique and specific embouchure. According to the different mouthpiece on each instrument there is a particular relationship in each embouchure involving the palate, lips, tongue, teeth and orofacial muscles. Musicians in general can be affected by soft tissue trauma in the course of playing the wind instrument. In saxophone and clarinet players the instrument is placed intra-orally with the single reed mouthpiece. The lower lip is retruded between the lower surface of the mouthpiece and the mandibular incisors. The saxophone and clarinet players usually use a folded paper to prevent pain and ulceration. **Aim:** To develop a simplified technique using a thermoforming material for fabricating a custom mouthpiece for wind instruments. Due to the fact that there is a different relationship between the oral cavity and the single-reed instruments, the double-reed instruments, and brass instruments, this research intends to evaluate the specific orofacial needs of wind instruments students from a Professional Music School in Oporto, (Escola Superior de Música e Artes do Espectáculo). **Materials and Methods:** The maxillary and mandibular casts were obtained after dental impressions of the wind instruments players. A thermoplastic material of ethylene vinyl acetate copolymer (EVA) of 1 mm thick was placed over the mandibular cast, which was taken to a vacuum machine, Bio-Estar. The new device obtained was cut at the edges of both canines. This new custom mouthpiece was limited to the anterior incisors to reduce the space and volume occupied intra-orally. A questionnaire was answered by the musicians regarding to their general oral health issues and related problems of their embouchure. **Results and Discussion:** There are specific orofacial considerations to each instrument groups. Usually there is pain associated by an irregularity or sharpness of the lower anterior teeth into the lip when playing in single reed-instruments (saxophone and clarinet). In brass players (trumpet) there can be discomfort and pain due to the pressure of the metal mouthpiece when the lips are pressed against rotated teeth or with protruded teeth. The development of the new custom mouthpiece had general acceptance in the student population of the Music School (E.S.M.A.E.), having a high effectiveness in saxophone and clarinet players. This appliance may change the traditional folded sheet of paper which was a “bulky” situation to a device that has retention. The process of fabrication of the custom mouthpiece for wind instruments is relatively simple, not very expensive and the musician refers that no longer “spits” paper at the end of a concert. **Conclusions:** It is important that Dentists have a general knowledge of the specificity of each wind instruments and it's relation with their oral health. Providing information that is relevant for musicians and developing medical devices that can be used to optimize their performance may be detrimental to their careers.

1 INTRODUCTION

A musical performance depends on high-level creative skills ranging from technical control of the instrument, to expressive and interpretative awareness and ability (Sloboda & Howe, 1994). However, there are other performance-related skills such as the ability to behave with 'correct' socio-cultural etiquette and the ability to cope with the potential stresses and demands of different venues and types of audience, that contribute to the success in performance (Davidson, 1997; Kokotsaki & Davidson; 2003).

Current research trends on music performance indicate that it is based on the knowledge musicians acquire through a great deal of deliberate practice to obtain high levels of skills (Ericsson, Krampe & Tesch Roemer, 1993; Sloboda & Howe, 1994; Davidson et al., 1997; Hallam, 1997; Clarke, 2002, Williamon, 2004). In fact, together with an early involvement, deliberate practice is one of the main convergence points to achieve excellence in any specific domain, be it sports or arts (Araújo et al., 2007; Sloboda & Howe, 1994).

Due to the great amount of practice undertaken by musicians, they have to deal with the physical constraints derived from the repetitive use of the instrument. For instance, the incidence of focal dystonia may be as high as one in 200 professional musicians (Altenmueller, 2000; Schuele et al., 2005) and research has shown several attempts to adequately retrain the dystonic musician (such as deLisle et al., 2007). *Pain* affects a large number of musicians. On a study that involved 57 orchestras worldwide James (1997) results indicated that 56% of the musicians involved suffered pain during the previous year, 19% reported strong pains that negatively affected the quality of their performance and forced them to stop playing. Research has also shown the musculoskeletal system to be the most frequently involved area of impairment (Morse et al., 2000 and Roset-lobet et al., 2000).

Orofacial problems are also included as result of the musicians' professional activity. Musicians start playing a specific instrument usually before they reach the age of ten years old. At this point of their growth the development of the roots of permanent teeth are not complete. Orthodontic problems, focal dystonia, herpes labialis, dry mouth, and temporomandibular joint disorders (TMD) can be identified as orofacial problems of career musicians. The most prevalent problems affecting wind players involve overuse of muscles resulting from repetitive movements of playing, especially because it requires increased ventilation and increased orofacial muscle activity.

Among the different type of wind instruments there is a specific muscular pattern to form the

embouchure. It is important that the dentists understand the increased orofacial activity applied by the musician when playing the musical instrument. There are different kinds of dentitions between wind players that can compromise the embouchure, for example crowded and rotated mandibular incisors will have an influence on the lower lip due to the pressure of a trombone. Likewise anterior sharp teeth can cause general discomfort and soft tissue trauma while playing single reed instruments (saxophone and clarinet).

It is noteworthy that James's (1997) results indicated that 83% of the musicians felt that their conservatoires or music colleges did not prepare them adequately for the physical requirements of their profession.

From the above-mentioned studies it seems clear, therefore, that (1) there is a great scope for engaging in interdisciplinary research that involves musicians, music educators, and in this case, dentists that can develop biomaterial devices with the intention, that they can eventually be used as a tool for an early intervention and prevention.

Hence the aim of the present work: (1) to contribute to the literature on musicians and well-being, as well as (2) to draw an intervention plan of oral health and well-being on a music college.

2 METHOD AND MATERIALS

As this study tends to explore the conditions of their oral health in relation to performance situations a questionnaire was addressed to 45 wind instruments students of Superior School of Music, Porto Polytechnic Institute. This questionnaire addressed their practice behaviours, the starting age as music student, the type of instrument they play, the number of hours of their practice activities, orofacial region affected in their performance preparation or public performances and specifically in there dentition if there was any tooth or teeth that felt a particular discomfort or pain with performance related tasks (Figures 1 and 2).

There were several reasons for the choice of this venue. Firstly, the main goals of a music college are artistic training. This is done in accordance with the *state-of-the-art* concepts of an ideal performance and a music college is expected to evolve these concepts when training future musicians. Secondly, a Music College is a place where there is a large number of people in the same situation, thus facilitating the process of completion and analysis of questionnaires.

This multi-disciplinary project had the intention of including receptive students in the development of a new custom mouthpiece for wind instrumentists. The maxillary and mandibular casts were

Diagnóstico e Tecnologia de Biomateriais III
TEMA - Avaliação do potencial uso de Lip Pressure Appliance (LPA) em instrumentos de sopro

**Universidade do Estado do Rio de Janeiro**
Instituto de Física
Departamento de Física Experimental

QUESTIONÁRIO

Nome: _____ Data: 11/04/2020
Idade: _____ Data de Nascimento: _____
Título e local de trabalho: _____
Estado civil: _____ R.º de Filhas: _____
Telefone: _____ e-mail: _____

1. Instrumento Musical:
Saxofone

2. Idade de início na atividade musical:
14

3. Formação Musical:
☒ Escola Profissional
☐ Conservatório:
☐ Ensino Supletivo
☐ Ensino Artístico
☐ Academia
☐ Bandas
☐ Escola de Música
☐ Professor Particular

4. Número de horas de prática musical por dia:
7-12 anos
13-18 anos
>18 anos

5. Posição que adota para tocar:
☒ Sentado
☐ De pé

6. Há 30 anos outro instrumento musical?
☐ Sim
☒ Não

7. Se sim, qual foi aquele?
13 anos
6.2 Durante quanto tempo?
7 anos

7. Tocava este instrumento ao mesmo tempo que o instrumento principal?
☐ Sim
☒ Não

8. Qual é a sua aperta principal (o que mais incomoda neste momento)?

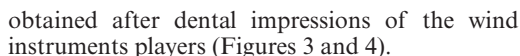
9. O início dos sintomas foi:
☐ Acontecendo aos poucos
☐ Repentino
☐ Outros: _____

10. O que poderia fazer, que não faz, para otimizar a sua performance musical?

11. Hábitos tabagísticos:
☐ nunca fumou
☐ ex-fumador há mais de 5 anos
☒ ex-fumador há menos de 5 anos
☐ fumador < 10 cigarros/dia
☐ fumador > 10 cigarros/dia

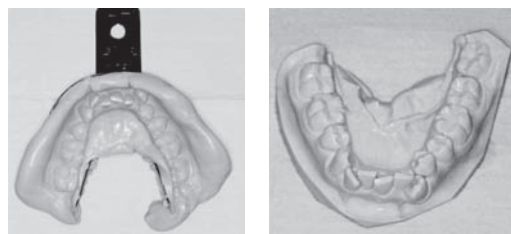
12. Hábitos de higiene oral:
12.1 Tipo de escova:
☐ elétrica
☒ manual
12.2 Número de escovas/dia:
☐ 0
☐ 1
☒ 2
☐ 3
12.3 Métodos auxiliares de higiene oral:
☐ fio dental
☐ bochecho com enxágue
☐ escovinha
☐ nenhum

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A thermoplastic material of ethylene vinyl acetate copolymer (EVA) of 1 mm thick was placed over the mandibular cast, which was taken to a vacuum machine. Bio-Estar (Figure 5).

This custom mouthpiece was manufactured from a relatively flexible thermoplastic material and configured and fitted to the lower dental arch of the player who uses it (Figure 6). This new device obtained was cut at the edges of both canines.



Figures 3 and 4. Impression and mandibular dental cast.

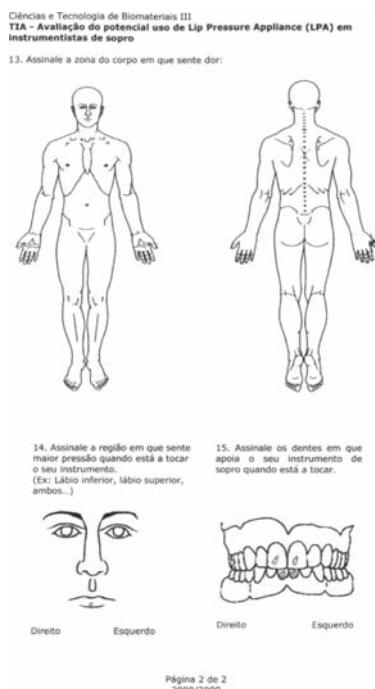


Figure 5. Bio-Estar machine were mandibular dental cast was placed with a thermoplastic material of ethylene vinyl acetate copolymer (EVA) of 1 mm thick, being previously heated and adapted with a pressure of 5 bars.



Figure 6. Custom mouthpiece for wind instrumentists in dental cast.



Figure 7. Custom mouthpiece view, intra-orally.

This new custom mouthpiece can be limited to the anterior incisors to reduce the space and volume occupied intra-orally regarding the specific needs of each particular wind instrumentist (Figure 7).

3 RESULTS AND DISCUSSION

3.1 *Studying musi – starting age*

As it could be seen in Figure 8 above, 83% of the subjects started playing an instrument before the age of twelve and 16% after the age of thirteen. It is also noteworthy that only 2% of the subjects started playing their instruments after the age of sixteen. These results are in line with Sloboda & Howe and Araújo et al., who claimed the need for an early involvement in such a high-achieving context as that of musical activity. In fact 43% of our sample has started playing before ten years—old and 40% between eleven and twelve years—old. There might be an explanation for this latter point. Music is not mandatory in the Portuguese National Curriculum until the fifth grade (age ten and eleven). In this way, it may well be that students in this group had both a slightly later first contact with music and therefore a slightly later decision to play an instrument.

3.2 *Age and practice time*

The time spent practicing the instrument evolved with the years of instruction and consequently the age of the subjects. As can be seen in Figure 9 below, it varied from one hour per day at the earlier stages of instruction (between seven and twelve years—old) to two to four hours between thirteen and eighteen years—old and four hours a day at the age of eighteen.

In this way, our findings largely confirm the great deal of deliberate practice necessary to undertake, in order to obtain a high—level skills (Ericsson, Krampe & Tesch Roemer, 1993), and

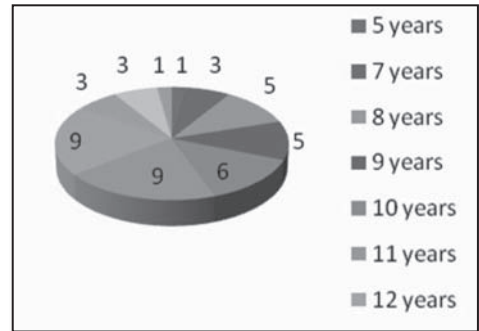


Figure 8. Beginning of musical activity age.

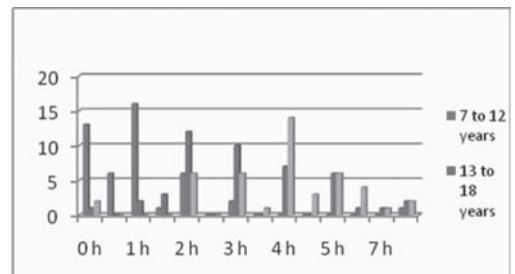


Figure 9. Number of daily practice hours.

more specifically, the great amount of practice involved in the development of musical expertise. (Sloboda & Howe, 1994; Davidson et al., 1997; Hallam, 1997; Clarke, 2002; Williamon, 2004). Given the above-mentioned early involvement in music, the amount of time spent practicing, and the physical demands of the music practice it would be expected that injuries, especially those like repetitive strain injury, would occur. This led to the next topic under discussion: which parts of the body were most affected, as far as injuries were concerned.

3.3 *Affected areas of impairment*

According to Morse et al., Roset-lobet et al., and Steinmetz the musculoskeletal system is the most affected area of impairment. As it is shown in Figure 4, the same findings were replicated in our study. This is particularly relevant for the high percentage of subjects that referred the the latissimus dorsi muscle as a highly affected region of the body. The column, column lombard, and trapezius were the next most frequently cited are of impairment followed by the forearm and the biceps muscle. This may well happen due to the huge physical demands of music performance, which requires wind instrumentalists to hold their instruments during the

great deal of practice undertaken. Besides having to play during long hours in unnatural postural positions, musicians have yet to achieve an excellent manual dexterity. This is done by moving the fingers and pressing the instrument's keys (one movement for each note) in a succession of hundreds of thousands (if not millions) of fine-coordination movements in a four hours practice session. This is certainly the reason why the wrist and thumb are also mentioned as affected areas of impairment. Results are shown in Figure 10 below.

3.4 Affected orofacial areas of impairment

The quality of both the vibration of the lips and the formed embouchure has a direct impact on the sound production of wind instruments. Consequently different types of wind instrumentalists need a high degree of deliberate practice in order to acquire the specific muscular pattern to form the required embouchure. This creates an increased orofacial activity. Perhaps this is the reason why the lips (particularly the inferior lips) emerged as the most affected facial area of impairment. This can be seen in Figure 11 below.

3.5 Affected dental areas of impairment

As it was shown in Figure 12, the most affected dental areas of impairment were the incisors, particularly the inferior. This result was somehow expected, due to the fact of the high percentage of the wind instrumentists students that participated on this questionnaire that play saxophone and clarinet Figure 13.

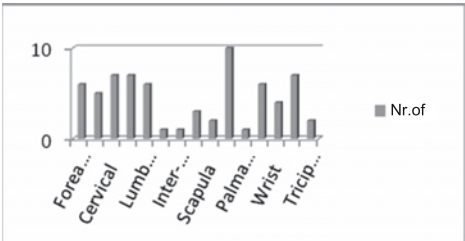


Figure 10. Affected body region.

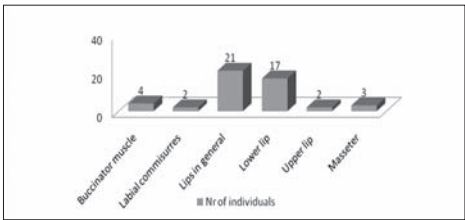


Figure 11. Orofacial affected region.

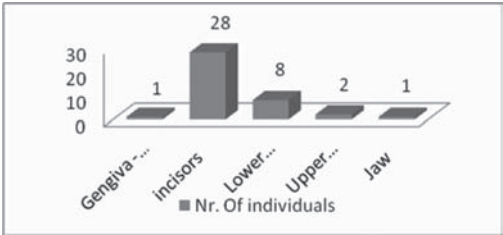


Figure 12. Affected dental region.

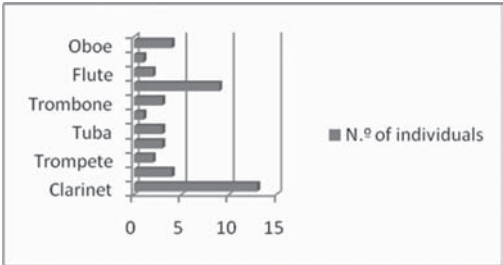


Figure 13. Instruments played by the 45 students of the Superior School of Music, Porto Polytechnic Institute.

There is a specific orofacial muscular activity, position and interrelationship of the mandible concerning the maxilla and finally the influence and position of the lips during the embouchure of the different wind instrumentists. As an example a saxophone and a clarinet have the intrusion of the lower lip which is placed between the lower surface of the mouthpiece and the mandibular incisal edges, with the fact that during the musical performance there is a natural increased pressure. In a certain mode this will explain the higher incidence of the incisors, more specifically the lower incisors. Regarding the brass instruments like the tuba, trombone, trompete, horn and trombone base they are placed extra-orally by pursing the lips against the metal cup-shaped mouthpiece. Once more the incisors have an increased pressure during the musical learning and performing. These results are able to understand the variety of needs wind instrumentists have.

The development of the custom mouthpiece for saxophone and clarinet players can mean a beginning of a strategic plan to establish in the future with performing arts, and the dental field. There are more questions raised at the end of this work, like which specific muscles act during their performance, will this custom mouthpiece be useful for every wind instrumentist or only the ones who used previously a folded sheet of paper over the lower incisal edge Figure 14.

Many devices were fabricated and applied to the wind instruments students (only clarinet and

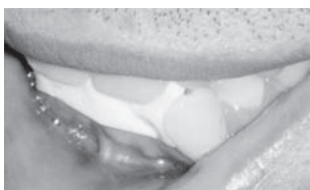


Figure 14. A folded sheet of paper that musicians found as a solution to prevent soft tissue trauma.

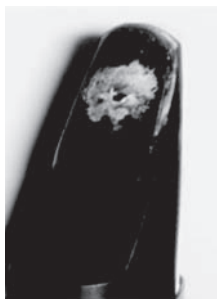


Figure 15. Mouthpiece of a clarinet player, it is possible to see the marks of the maxillary central incisors.

saxophone players due to their embouchure). They found them comfortable and particular interesting in the point of view that they no longer ended a concert spitting paper (of the protection they used before) that was destroyed by the saliva. Having the perception that the sound quality was exactly the same requires more time of adaptation to the custom mouthpiece, but there were students who said that no difference was noticed. Once more this brings the idea of a new future studies on noticing any kind of difference in the final result of music sound quality with the use or not of this medical device.

The process involved in the elaboration of the custom mouthpiece for wind instruments is relatively simple and economic. The rotation of a mandibular incisor or a sharp edge should not be eliminated by the reduction or remodeling of tooth surfaces with irreversible procedures. A conservative option with the development of this custom mouthpiece for wind instruments can have a high impact on students oral health, as well as to enhance perceptions of success and wellbeing in relation to performance situations were wind instrumentists actually apply high forces with their teeth, Figure 15.

4 CONCLUSION

Given the above results, it is surprising that musician's feel their conservatoire did not prepare them adequately as James's study has pointed out. Therefore, musicians' health and well-being should

become an essential element within the training of musicians. Hopefully an interdisciplinary approach such as the one carried out in the present work may contribute to further the understanding and assisting of musicians on their primary role: to make music. It is our belief that educational institutions have the responsibility to carry out research in the area, and, ideally, to provide music colleges and conservatoires advice on how to effectively promote the health of their students. They will be the future musicians and music educators.

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Minor maxillofacial bone augmentation for dental prostheses implantation

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ABSTRACT: Dental implantation is a common procedure in dentistry, and although several factors such as implant design, implant surface, surgical technique, bone type and loading conditions have been shown to influence implant-bone response, the insufficient bone volume in the implantation site is a serious limitation to a good outcome of the procedure.

In this study, a patient with the early removal of two opposite molars is considered. A virtual prototyping model was built to characterize the dimension and the morphology of the patient's maxillofacial structure. This anatomical model was based on hospital scanning data in order to permit an effective reconstruction planning and to infer about the implant's behavior and stability, considering key aspects like implant diameter and length.

1 INTRODUCTION

1.1 Maxillofacial bone and implantology

Implantology is a well developed clinical field and dental implant procedures are part of the current standard dentistry techniques. For aesthetic as well as pathological reasons, bone and soft tissue regeneration or augmentation are frequently performed.

Specifically, patients who suffered from early teeth removal or tumour resection, and patients with systemic diseases like diabetes or osteoporosis have a significant bone mass reduction of the alveolar ridge bone, both in terms of height and in terms of thickness (Nikolidakis et al. 2009). This reduced bone mass, if less significant, forces the surgeon to use shorter height implants, which usually implies implant stability and the success rate of implantation will be reduced (Hupp et al. 2008). In most cases, however, in the case of a significant bone loss, the implantation without previous bone augmentation is altogether unfeasible. In such cases, several techniques are presently in use in clinical dentistry. Each technique has its own advantages and limitations and the choice of the technique used for each patient depends on the particular clinical case and relies mainly on the clinician. These techniques will now be briefly presented.

1.2 Maxillofacial bone augmentation technique

1.2.1 Osteocondensation and alveolar crest splitting

The main characteristic of the osteocondensation and alveolar crest splitting techniques is that they require no bone graft or scaffold to perform the bone augmentation procedure. Osteocondensation relies on the condensation of bone in the site where the implant is to be placed by using specific tools that induce pressure on the bone, and is mostly used in the maxilla. Alveolar crest splitting is used to widen the alveolar ridge and is performed by producing a green stick fracture at the base of the alveolus (Sándor et al. 2003). These techniques are widely used for bone augmentation, since they reduce reconstructive morbidity. However, they have important limitations which include resorption of the augmented bone some time after the procedure and poor reproducibility when simultaneous bone augmentation and implantation is performed (Funaki et al. 2008).

1.2.2 Bone grafts onlay and inlay

Currently, the most often used bone augmentation procedure used is the placement of bone grafts on or in the implantation area. Bone grafts can come from autologous, allogeneic or xenogeneic sources, where autologous are the preferred sources when

their use is possible. In both the onlay and inlay techniques, blocks of bone (organic and cellular components are removed) are fixated to the native bone with rigid screws, the difference being that with the first method, the bone graft is fixated on top of the atrophic alveolar crest and with the second, the bone is placed in a cavity created in the atrophic bone (Hupp et al. 2008). Positive clinical results are usually obtained with this technique. Nevertheless, resorption of the inorganic matrix of the bone graft is usually very slow and fibrous encapsulation might occur, which frequently leads to implant failure.

1.2.3 Sinus lift

This technique is used for augmentation of the maxillary alveolar ridges. It is performed by placing bone graft material inside the sinus, and is usually performed by the lateral window technique or the sinus floor elevation technique (Jurisic et al. 2008). It can be performed if severe bone loss has occurred, although recuperation time is long and varies for each patient.

1.2.4 Alveolar distraction osteogenesis

The alveolar distraction osteogenesis is a technique used to lengthen osteotomized bone. A gap is created and an apparatus is placed, which allows the surgeon to slowly move the osteotomized segments of bone, to allow new bone formation. The daily rate of alveolar crest distraction is 0.25–0.5 mm, which allows a bone gain of 4 to 7 mm in 30 days (Sándor et al. 2003). Although this procedure has reduced reconstructive morbidity, since no bone grafts are harvested, it requires daily adjustments of the distraction device and a second surgery to remove the device (Saulacic et al. 2009). This eventually leads the patients to prefer another, less uncomfortable solution.

1.2.5 Guided bone regeneration

Guided bone regeneration is a technique intended to prevent the invasion of the periodontal wound by fibrous connective tissue and epithelium, since this phenomena prevents bone regeneration or augmentation. It consists of placing a membrane, non-resorbable or resorbable, over the bone defect, which will work as a barrier and allow the underlying bone to regenerate. Good results have been presented for augmentation with this procedure. Its main limitations are the need of a second surgery to remove the membrane (for non-resorbable membranes), possible exposure and infection (Kinoshita et al. 2008).

2 CASE STUDY

In this study, an interesting clinical case will be studied which allows to perform two different

techniques on two different implant sites with similar bone loss and morphology. This particular arrangement makes it possible to perform direct comparison of the outcome of the two techniques chosen.

The subsequent follow up of bone mass regeneration and augmentation by imaging techniques will produce enough data to model and perform computational simulation the process of bone augmentation for each technique.

A first approach on the construction of the computational anatomical model was carried-out considering a bi-dimensional cross-section model of the critical section of the implant site, as it allows predicting the stability of the selected implant in terms of implant size and geometry.

2.1 Description of clinical case and techniques

Initially one male patient, 37 years old, is scheduled for bone augmentation of the alveolar ridge and subsequent placement of molar implants in the maxilla.

The patient had both molars number 19 and number 30 removed 20 years previously and since no implants were placed for a long period of time, significant bone resorption occurred in the alveolar crest. As can be seen in Figures 1–2, the patient has significant loss of bone height and thickness at the tooth removal sites.

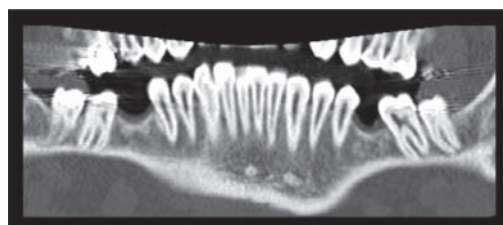


Figure 1. X-Ray scan of the clinical case.

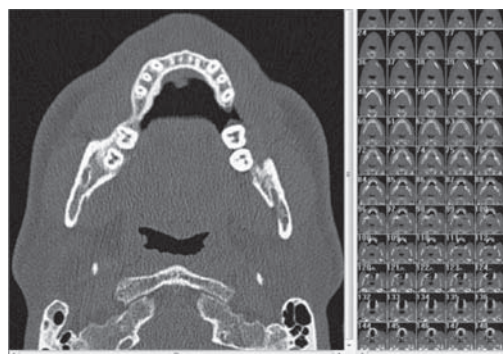


Figure 2. CT scan of the clinical case.

Implants placed will be of the Brånemark System® (NobelBiocare™ AG). Two possible implants can be used: the Brånemark System® Mk III Groovy for narrow platforms with 8.5 mm height (Figure 3) or the same model for narrow platforms but with 5.5 mm height (Figure 4).

The patient's alveolar bone height is insufficient to place the longer implant directly, without performing bone augmentation. The placement of the shorter implant eliminates the need for bone augmentation but reduces implant stability and the probability of successful implant osseointegration. Therefore, a bone augmentation of approximately 3 mm was considered advisable by the clinician, to render the placement of the longer implant possible.

According to the clinical case presented, the surgeon proposed three possible bone augmentation procedures to be used: guided bone regeneration, autologous bone grafting and alveolar

distraction osteogenesis. Of the three methods possible, distraction osteogenesis is the less recommended due to the intensive nature of the procedure and the discomfort caused in the patient. It will only be used if none of the previous methods are feasible or possible to be performed with the patient's consent.

Guided bone regeneration and bone grafting are the preferred methods. Guided bone regeneration will be performed with a synthetic scaffold (Bio-Oss®, Geistlich Biomaterials, Geistlich Pharma AG) enveloped with a biodegradable membrane (Bio-Gide®, Geistlich Biomaterials, Geistlich Pharma AG), to eliminate the need of a second surgery.

Autologous bone grafting will be performed by the onlay technique. An initial surgery will be performed to excise a disc of bone from the mandible.

This disc will then be superimposed on a previously prepared bone bed in the defect site. The final choice of which techniques will be used will rely on the patient but two possibilities are the most probable: 1) Autologous bone grafting on one implant site and guided bone regeneration on the other; 2) Placement of shorter implant on one site and guided bone regeneration on the other.

3 STRUCTURAL STUDY

In this study, a bi-dimensional cross-section computational model of both Brånemark System® implants, as well as the predicted anatomical location was built in order to infer about their predicted stability in terms of critical parameters, such as implant length and implant diameter.

3.1 Introduction

The analysis of implant's structural performance requires the development of numerical models. Finite element method is widely used to investigate the mechanical behavior of muscular-skeletal systems in order to assess strain and stress distribution within different anatomical structures, particularly, around biomedical implants, and to explore the influence of these distributions on their long-term stability.

Furthermore, finite element models allow the optimization and customization of design geometry of an artificial bone substitute, as well as the best placement for fixation and the selection of the most suitable materials for a given application (Zannoni et al. 1998). Many pathologies (Viceconti et al. 2004, Rietbergen et al. 2002, Kopperdahl et al. 2002) or simply ageing (Testi et al. 1999) can also make use of these models.

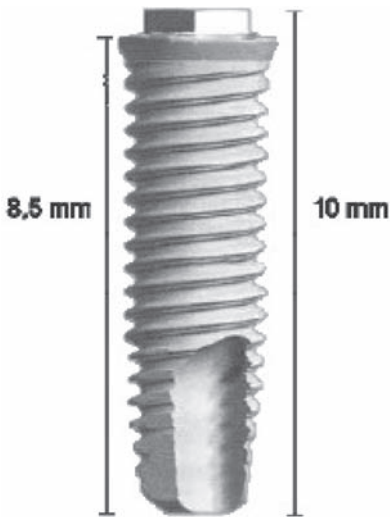


Figure 3. Brånemark System® Mk III Groovy implant for narrow platforms with 8.5 mm height.

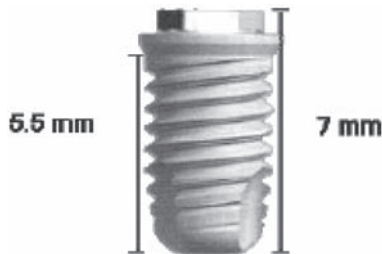


Figure 4. Brånemark System® Mk III Groovy implant for narrow platforms with 5.5 mm height.

However, the development of finite element models of human bones is difficult due to their 3D complex geometry, due to the nonhomogeneous (with cortical and cancellous regions) and orthotropic properties and due to the complex and variable loading.

Most of the finite element models of bones used in orthopaedic biomechanics research are based on generic anatomies (Viceconti, 2004). The analysis of real geometries is a step in front, as it provides tailor-made solutions. Although modern 3-D imaging modalities, like computed tomography, magnetic resonance imaging or ultrasound imaging are widely applied in current medical practice, it is not easy to obtain robust vectorial 3-D models to be used in finite element method.

Bones are constituted of cancellous regions at the extremities and a cortical region in their centre, with quite distinct properties. Therefore, it is necessary for finite element modeling to distinguish the two types of material areas.

Different parts must also be considered in the computed model of bone to allow an accurate modeling of its global mechanical behavior. In fact, it is accepted that bone has an anisotropic behavior, and so any attempt at modelling with isotropic constitutive properties is an approximation to the complexities of the real material.

The numerical problems of anisotropic simulation of bone modeling arise from the lack of a comprehensive data bank incorporating the material properties of bone as a function of the orthotropic load directions.

Currently CAD modeling software has automatic mesh generators incorporated which simplifies

notably mesh definition. The user only controls the average size of finite elements, although local refinements can be defined. Ideally the mesher should be able to provide a reasonable mesh without any user input regarding mesh size.

This approach is widely used considering its simplicity. Linear or quadratic tetrahedral elements are usually used in automatic meshing. A great number of elements can be easily attained, implying a great numerical effort. Contact option is usually available and regions with different properties can also be defined. On the other hand, when the structure has regions with different materials properties, a great mesh control is required, therefore structured and parametric meshes are recommended.

3.2 Finite element model

A simplified 2D model was developed to understand the structural performance of the implant, and particularly the effect of the diameter and length of the implant. The critical section on the mandible was assessed directly from the CT acquired data (Figure 6) and the corresponding contour was evaluated and modelled in a CAD software suite. Implant contour was also derived considering their correspondent geometry and dimensions. A thickness of 1 mm was considered for the 2D layer studied, which replicates the local geometry of patient's jaw in the region where the molar is missing. Figure 7a shows the geometry considered for the 2D layer. Different regions were defined considering the different bone type. A layer of about 1 mm was considered at surface, corresponding to cortical bone. This material was

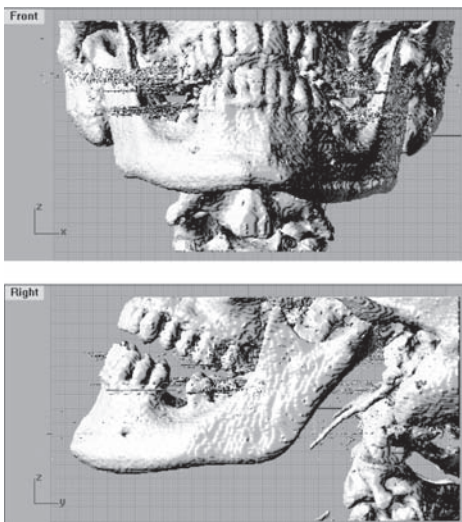


Figure 5. 3-D models of the clinical case.

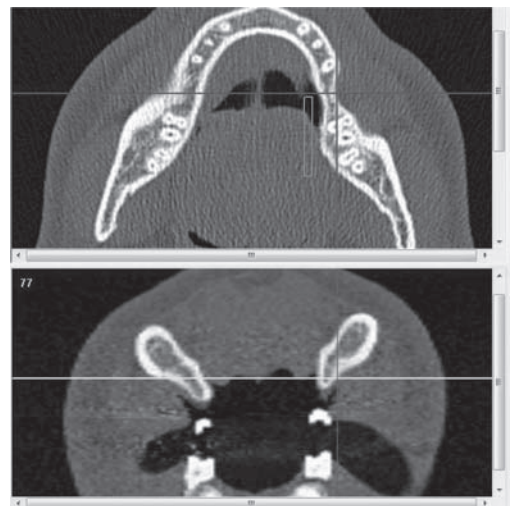


Figure 6. Contour assessment of the critical section of the patient's mandible.

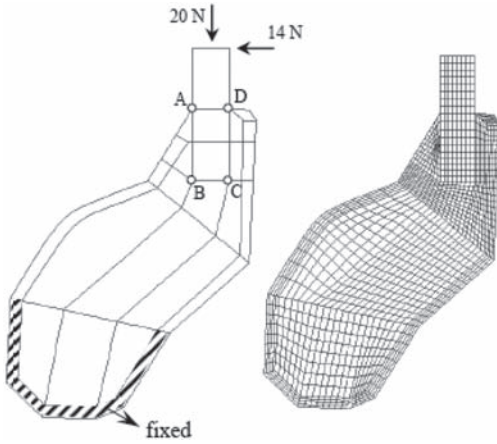


Figure 7. a) Geometry and b) finite element mesh.

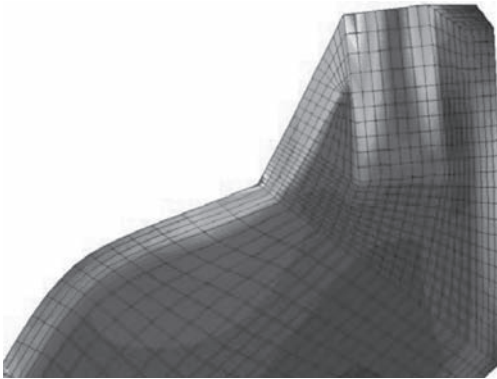


Figure 8. Equivalent von Mises stress field ($d = 3$ mm).

assumed to be homogeneous, isotropic and linear elastic with $E = 18$ GPa and $\nu = 0.3$ (Wirtz et al., 2000), being E the Young's modulus and ν the Poisson's ratio. Inside the 2D layer, the cancellous bone was also assumed to be homogeneous, isotropic and linear elastic but with $E = 5$ GPa, $\nu = 0.12$.

For the implant the elastic properties assumed were $E = 100$ GPa, $\nu = 0.3$. Forces of 14 and 20 N were applied at the top of implant along horizontal and vertical directions, respectively, as indicated in Figure 7a. The layer was fixed at the lower region, as illustrated.

Figure 7b shows the finite element mesh considered in the discretization. It is composed of 1384 4-node linear isoparametric elements and 1823 nodes. Figure 8 shows the stress field. The most rigid parts, i.e., the cortical bone and implant, have higher stresses levels, as could be expected.

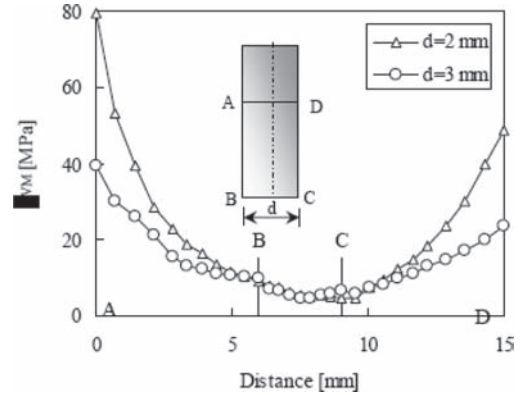


Figure 9. Equivalent von Mises stress versus distance to point A.

The cortical bone is the main responsible for the stability of the implant in the model developed. Figure 9 presents the equivalent von Mises stress along the periphery of the implant, i.e., from point A through points B and C up to point D (Figure 7a). Peak values were obtained at positions A and D, but the highest value is at point A. The effect of diameter (d) and length of the implant was studied. The effect of d is presented in Figure 9.

The decrease of the diameter of the implant increases significantly the stress level. On the other hand, the length of implant was found to have a minor influence on the stress level. This could be expected considering that the cancellous bone has a minor influence on the global rigidity.

According to the results obtained, the stability of the implant depends mainly on its diameter, and is not significantly influenced by its length. Notice that the numerical model assumes a perfect connection between the bone and the implant, which is not correct, particularly at the beginning of dental implantation.

Further work is required to build a 3D model that replicates effectively the real geometry and to improve the bone/implant contact model.

4 CONCLUSIONS

This study focused on a clinical case regarding a patient with early removal of two opposite molars. As more than 20 years have passed without implantation, significant bone resorption has taken place so that bone augmentation procedures had to be analyzed and discussed.

Alternatively, considering the patient's reduced alveolar bone height, shorter implants were selected

and their stability in the anatomical site was analyzed.

Based on hospital scanning data, a virtual prototyping model was built to characterize the dimension and the morphology of the patient's maxillofacial structure.

The implant's behavior was simulated and analyzed, considering key aspects like implant diameter and length. According to the results obtained, the stability of the implant depended mainly on its diameter, and was not significantly influenced by its length. However, the numerical model assumed a perfect connection between the bone and the implant, which is not correct, particularly at the beginning of dental implantation.

Further work is required to build a 3D model that replicates effectively the real geometry and to improve the bone/implant contact model.

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